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PhD THESIS

HUMAN LOCOMOTION:
CENTRE OF MASS AND SYMMETRY

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GLOSSARY

Chapter 1

BCOM = centre of mass of human body

ADL = activity daily life

EMG = electromyogram

GC = gait cycle

COG = centre of gravity

COP = centre of pressure

COM = centre of mass

SI = System International

m = mass [kg]

ρ = density [kg/m^3]

V = volume [m^3]

M = total mass of a body or a system [kg]

r_{BCOM} = three-dimensional position vector of the BCOM [m]

x_{BCOM} = forward position vector of the BCOM [m]

y_{BCOM} = vertical position vector of the BCOM [m]

z_{BCOM} = lateral position vector of the BCOM [m]

F = force exerted on a body or a system [N]

a = linear acceleration of a body or a system [m/s^2]

3D = three-dimensional

TE_{BCOM} = total energy of the BCOM [J]

KE = kinetic energy [J]

PE = gravitational potential energy [J]

TE = total energy [J]

EL or $U\varepsilon$ = elastic energy [J]

Chapter 4

MoCap = motion capture

IR = infrared cameras

Chapter 5

S.D. = standard deviation

Chapter 6

n = number of stride cycles

Ay_0 = vertical position of the BCOM [m]

cos = cosine function

sin = sine function

φ = phase angle or phase coefficient [rad]

arcsin = arcsine function

tan = tangent function

$y(t)$ = periodic function

t = time period [s]

T = period [s]

a_0 = constant term

a_i = cosine Fourier Series coefficients

b_i = sine Fourier Series coefficients

a_1 = fundamental cosine frequency coefficients

b_1 = fundamental sine frequency coefficients

a_n = *n* derivative cosine frequency coefficients

b_n = *n* derivative sine frequency coefficients

a = amplitude quantifying the magnitude of the oscillations

f = frequency of the oscillations

C_0 = equivalent of a_0 (= the eventual constant term)

C_n = *n* amplitude coefficient harmonics

φ_n = *n* phase coefficient harmonics

A = amplitude coefficient

φ = phase coefficient

Chapter 7

DLS = Digital Locomotory Signature

Appendix 7.2

MAX = Maximal value [m]

Min = minimal value [m]

Chapter 8

SI = Symmetry Index

Chapter 9

r = mean vector length

s^2 = angular variance

s = angular standard deviation

Chapter 10

SF = stride frequency [Hz]

SL = stride length [m]

DF = duty factor [%]

W_{ext} = mechanical external work *per* unit mass and distance [J/(kg·m)]

W_v = absolute work to lift the BCOM [J/(kg·m)]

W_f = absolute work to accelerate the BCOM forward [J/(kg·m)]

R = energy recovery percentage [%]

W_{int} = mechanical internal work *per* unit mass and distance [J/(kg·m)]

W_{tot} = mechanical total work *per* unit mass and distance [J/(kg·m)]

Chapter 11

L = Lissajous

Chapter 12

MW_{int} = Measured mechanical internal work *per* unit mass and distance [J/(kg·m)]

PW_{int} = Predicted mechanical internal work *per* unit mass and distance [J/(kg·m)]

q = compound term accounting for the inertial properties of the limbs

a = fractional distance of the lower limb centre of mass to the proximal joint

g = average radius gyration of limbs, as a fraction of the limb length

b = length of the upper limb, as a fraction of the lower limb

mu* = fractional mass of the upper limb mass

ml* = fractional mass of the lower limb mass

s_{ST} = the average speed, relative to the BCOM, of the foot when in contact with the ground (stance) [m/s]

s_{SW} = the average speed, relative to the BCOM, of the foot during the swing [m/s]

Chapter 14

OR = occasional runners

SR = skilled runners

TR = top runners

Chapter 15

RF = radiofrequency fields

NMR = Nuclear Magnetic Resonance

CT = Computed Tomography

MRI = Magnetic Resonance Imaging

I = non-zero spin [$kg \cdot m^2/s$]

μ = non-zero magnetic momentum [$A \cdot m^2$]

γ = gyro-magnetic ratio [$A \cdot m^2$]

ω_0 = precessional frequency [MHz/T]

B_0 = external magnetic field strength [Tesla]

SAR = specific absorption rate

Chapter 17

SSR = Static Symmetry Ratio

CV = coefficient of variation [%]

r = cross-correlation coefficient

d = displacement of a region in respect to the other

C = cost of progression [J/(kg·m)]

$V'O_2$ = oxygen consumption [ml/(min·kg)]

$V'O_2rest$ = rest oxygen consumption [ml/(min·kg)]

Chapter 19

V'E = ventilation [L/min]

RE = respiratory equivalent [(ml/(min·kg))/(bpm)]

HR= heart rate [bpm]

RER = respiratory exchange ratio

Chapter 20

η = mechanical 'apparent' efficiency

ΔBCOM = spatial variability of the BCOM [m]

Δt = temporal variability of the BCOM [s]

M = mean of the variability of the BCOM

CV = coefficient of variation of the variability of the BCOM

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ENCLOSED CD CONTENTS

It was really difficult to insert all the data in the thesis due to the significant amount of information we recorded and presented. Consequently, we have decided to create a CD which contains the main results of both studies. All these topics have been widely investigated and discussed in each chapter of the thesis.

Therefore, this CD constitutes additional material which can be useful if someone, in the near future: a) wants to understand our data better; and b) wants to check our data to go on with biomechanical and physiological experiments. To be precise, the single material inserted in the CD has been arranged following the same distinction as proposed in the thesis. In order to make its reading and consultation easier, we want to briefly illustrate the main files contained in the CD.

Firstly, the CD has been divided in two parts corresponding to the first and the second study, respectively (FIRST STUDY and SECOND STUDY). Secondly, both projects have been further separated into single chapters.

As far as the first study is concerned:

- **Chapter 6** contains average anthropometric parameters of all ages; all files in *.bcm and *.res format; a comparison among different filter orders; all spreadsheets derived from the *Motion Filter Analysis* *.vi (in LabVIEW); in each testing condition, average range volumes, Froude numbers, Ay0 values and BCOM excursions;
- **Chapter 7** contains all templates (in both each subject and means of age groups), in both level and gradient gaits (males and females); graphical representation of 3D *contours* (in *Grapher*) as a function of age, speed and gradient (males and females); harmonic coefficients (and their pattern) as a function of age, Froude Number and speed in all testing conditions; results of the regression analysis for both average amplitude and phase coefficients; coefficients data used in statistical analysis;
- **Appendix 7.2** contains average BCOM excursions;
- **Chapter 8** contains average Symmetry Indexes (and mean overall Symmetry Indexes), in both level and gradient gaits; Symmetry Index data used in statistical analysis;
- **Chapter 9** contains amplitudes (with their linear standard deviations) and phases (with their circular standard deviations) in all testing conditions; graphical representation of both harmonic coefficients in polar logarithm graphs (in *Grapher*) as a function of speed and gradient (without and with standard deviations);

- **Chapter 10** contains average simple and complex biomechanical data, in both level and gradient gaits (males and females); biomechanical variables used in statistical analysis;
- **Chapter 11** contains average energy Lissajous template, in level gaits (males and females); comparison between discrete measurements and corresponding continuous mathematical functions;
- **Chapter 12** contains average spreadsheets of measured and predicted mechanical internal work, in both level and gradient gaits (males and females); average values of predicted internal work and term compound *q*.

As far as the second study is concerned:

- **Chapter 16** contains average anthropometric parameters of all runners;
- **Chapter 17** contains all spreadsheets derived from the *Motion Filter Analysis* *.vi (in LabVIEW); all templates in each recordings (in both each runner and means of groups) and average templates (independently of recordings); all files in *.bcm and *.res format;
- **Chapter 18** contains Static Symmetry Ratio (for three anatomical regions) values in all groups; Static Symmetry Ratio data used in statistical analysis; cross-correlation coefficients in all groups; graphical representation of 3D *contours* (in *Grapher*) as a function of running (occasional, skilled and top runners); harmonic coefficients (and their pattern) as a function of speed; average Symmetry Indexes (and mean overall Symmetry Indexes); Symmetry Index data used in statistical analysis; amplitudes (with their linear standard deviations) and phases (with their circular standard deviations); graphical representation of both harmonic coefficients in polar logarithm graphs (in *Grapher*) as a function of speed (without and with standard deviations); files derived from $K4b^2$ equipment; average values of metabolic cost;
- **Chapter 19** contains the main physiological variables in both treadmill and track condition;
- **Chapter 20** contains average simple and complex biomechanical data, in all runners; average values of mechanical 'apparent' efficiency; average energy Lissajous template; comparison between discrete measurements and corresponding continuous mathematical functions; single subject files in *Acqknowledge* software in both occasional and skilled runners; corresponding average absolute and relative variability of the BCOM.

SCHEDULE FOR RESEARCH ACTIVITY

A schedule is presented here below summarising the main phases of our research activity. As can be seen, each phase has been developed and carried out within a specific limited period.

Literature review period	January 2007 - June 2007
Analytical techniques assessment **Subject recruitment**	March 2007 - September 2009
Data collection	March 2007 - December 2009
First proposal presentation and approval to Doctorate School	13th June 2007
Data analysis	October 2007 - December 2009
International congress participation (POSTER) Evaluation of a rehabilitation protocol by repeated motion capture analysis after ACL reconstruction: a single subject study in rugby *Nardello F., Marcolin G., Petrone N.*	13th ECSS Congress 9 - 12th July 2008 Lisboa
Second proposal presentation and approval to Doctorate School	12th November 2008
International congress participation (POSTER) Mechanical efficiency and running economy in occasional and skilled runners and their relationship with the kinematic variability of the body centre of mass *Nardello F., Ghezzi S., Zamparo P., Ardigò L.P., Minetti A.E.*	14th ECSS Congress 24 - 27th June 2009 Oslo
International congress participation (ORAL) Human locomotion: right/left symmetry in 3D trajectory of the body centre of mass *Nardello F., Ardigò L.P., Minetti A.E.*	18th ESMAC Congress 17 - 19th September 2009 London
Paper submission to Human Movement Science Measured and predicted mechanical internal work in human locomotion *Nardello F., Ardigò L.P., Minetti A.E.*	First submission 8th September 2009 Second submission 13th February 2010
Third proposal presentation and approval to Doctorate School	16th December 2009
Writing PhD thesis	August 2008 - February 2010
Definitive target date for Thesis submission	28th February 2010

Table 1. Schedule for research activity.

PREFACE

The scientific study of human locomotion is of interest to both those in research laboratory and sport/clinical settings. Over the last few years, human movement studies have received new impulse from improved imaging technologies and highly sensitive tools (e.g. to measure forces/accelerations and so on). Nowadays, biomechanical analysis is able to approach both human and animal locomotion in perturbed/extreme conditions. Importantly, new gait analysis protocols have been introduced as evaluation tools both in sport (e.g. elite sportsmen) and in the clinical context (e.g. neurological patients).

In most cases, fields of interest are kinematic, kinetic and activation variables.

Although the technological achievements have spread the use of Biomechanics to sport and clinics, it is important to note that these approaches still suffer from intrinsic technical complication and lack of clear understanding of the significance of the data by the general user. As a result, in spite of the quantitative data that can be obtained by biomechanical approaches, it is the experienced eye that plays the most prominent role in prognostic and diagnostic processes in sport or in a clinical contest. Therefore, simplified and easy functional evaluation biomechanical protocols are strongly required.

In addition, nowadays, it becomes very important to develop a *multilateral approach*: both qualitative and quantitative. In fact, most of the biomechanical research concentrates on single joint dynamics, which is worthwhile and gives information on local changes or problems in its functional status. However, it is amazing to find that current literature lacks an accepted mathematical method which fully describes and quantifies the individual behaviour of the centre of mass of the human body (BCOM) during locomotion.

This physical imaginary point, the centre of mass of the human body, represents a relevant but often neglected gait analysis variable. By definition, in the BCOM, we can summarize the whole body movement and the translational vector for the momentum of the body mass. The three-dimensional (3D) trajectory of the BCOM could represent the individual signature as gender, age, gait of locomotion, speed and gradient vary. In this way, it makes clearly to take the BCOM as the most representative moving point in order to analyse its motion in various conditions of human locomotion.

Importantly, the main purposes of our research projects are:

a) to mathematically describe kinematic variables of the BCOM over space (e.g. primarily, left/right asymmetry) and time (e.g. when an anomaly is detected in the stride) domains.

Significant temporal and spatial parameters featuring the BCOM pattern will be evaluated. The mathematical representation will be used to investigate the changes due to gender, age, gait, speed and gradient;

b) to quantitatively describe the individual gait signature (the so-called *Digital Locomotory Signature*);

c) to assess, in each movement direction, the symmetry of BCOM trajectory between the two stride phases (the so-called *Symmetry Index*);

d) to build up an initial *normality* full-comprehensive database of reference equations (coefficients: amplitudes and phases) of the mathematical function representing the 3D displacement of the BCOM over many different healthy conditions. In such a way, impaired (degenerative, traumatic, aging pathologies) and improved (training, sport, active/passive aids) locomotion could be compared to healthy gaits. To be more precise, the variations will be quantified both in the temporal and spatial domains. In such a way, the quantitatively evaluation of the changes in the global locomotion pattern, during and following rehabilitation treatments and training sessions, could help to modulate the interventions and to better understand their overall effects;

e) to investigate the changes due to gender, age, gait, speed and gradient in main (simple and complex) biomechanical variables. These variables have been calculated by using a discrete method (cycle by cycle), a mathematical continuous function (based upon Fourier Series) and a model equation (indirect measurement);

f) to verify whether a significant relationship between static anatomical/kinematic functional symmetries and running economy exists in humans featuring varying running levels;

g) finally, to deeply investigate bioenergetics and biomechanics of running.

To achieve these aims, two different studies have been drawn up:

1. the **first project** (*Human locomotion: symmetry in the 3D trajectory of the body centre of mass*) seeks to apply both a mathematical method (function) and a valid evaluation protocol to explain the three-dimensional displacement of the BCOM over time (**points a - d**). The purpose of this research project, focusing on the BCOM as the investigation object fulfilling such a need, is achieved through an innovative use of classic biomechanical procedures, too. Preliminary data from sedentary subjects indicate that the methodology for the quantification of the 3D displacement of BCOM is sound and the development of the mathematical method and evaluation protocol looks promising.

Finally, this project will try to verify both discrete *versus* continuous functions and direct *versus* indirect measurements in analsying the pattern of biomechanical variables (**point e**);

2. the **second project** (*The relationship between bodily symmetries and running economy in humans*) will verify both static anatomical and kinematic functional symmetries as important and relevant indicators (determinants) of running economy (**point f**).

   Finally, this project will illustrate the main aspects of bioenergetics and biomechanics in runners with different trained levels (**point g**).

In order to present and fully describe these studies, the following sections have been arranged as follows:

- chapter 1 contains a general introduction focused on the body centre of mass as a useful gait analysis feature in human locomotion (in 'Introduction & background literature');
- chapter 2 contains a brief presentation of literature studies concerning the relationship between static/dynamic symmetries and running economy in both animal and human models (in 'Introduction& background literature').

1. 'Human locomotion: symmetry in the 3D trajectory of the body centre of mass'

In this first project:

- chapter 3 contains an accurately presentation of the main features/aspects of the study;
- chapter 4 accurately describes the instrumentation used to collect and process kinematic and biomechanical data (in Part 1: 'Materials and methods');
- chapter 5 presents methods, especially focusing on subjects and test protocols (in Part 1: 'Materials and methods');
- chapter 6 shows the mathematical method to analyse the three-dimensional displacement of the body centre of mass (in Part 1: 'Materials and methods');
- chapter 7 illustrates the main results concerning the Digital Locomotory Signature, and relative discussion. In the Appendix 7.1, results of statistical analysis in harmonic coefficients have been illustrated. Finally, in the Appendix 7.2, the average excursions of the BCOM have been presented and discussed (in Part 2: 'Results and discussion');
- chapter 8 illustrates the main results concerning the Symmetry Index, and relative discussion. In the Appendix 8.1, results of statistical analysis have been illustrated (in Part 2: 'Results and discussion');
- chapter 9 illustrates the main results concerning polar logarithm graphs, and relative discussion (in Part 2: 'Results and discussion');

- chapter 10 shows results concerning simple and complex biomechanical variables, and relative discussion. In the Appendix 10.1, results of statistical analysis have been illustrated (in Part 2: ‘Results and discussion’);
- chapter 11 uses Fourier analysis to calculate some complex biomechanical variables: external work and energy recovery percentage (in Part 2: ‘Results and discussion’);
- chapter 12 illustrates the application of a mathematical model equation to predict mechanical internal work (in Part 2: ‘Results and discussion’);
- chapter 13 contains conclusions, focusing both on peculiarities and limitations of the study (in Part 3: ‘Conclusions’).

2. ‘The relationship between bodily symmetries and running economy in humans’

In this second project:

- chapter 14 contains an accurately presentation of the main features/aspects of the study;
- chapter 15 accurately describes the instrumentation used to collect and process imaging, biomechanical and physiological data (in Part 1: ‘Materials and methods’);
- chapter 16 presents methods, especially focusing on subjects and test protocols (in Part 1: ‘Materials and methods’);
- chapter 17 shows methods used to analyse static symmetries, dynamic ones and running economy (in Part 1: ‘Materials and methods’);
- chapter 18 shows results about both static/dynamic symmetries and running economy, and relative discussion (in Part 2: ‘Results and discussion’);
- chapter 19 focuses on the bioenergetics of running on different surfaces (e.g. treadmill *versus* over-ground), and relative discussion (in Part 2: ‘Results and discussion’);
- chapter 20 focuses on biomechanical variables on treadmill running, and relative discussion (in Part 2: ‘Results and discussion’);
- chapter 21 contains conclusions, focusing both on peculiarities and limitations of the study (in Part 3: ‘Conclusions’).

INTRODUCTION

&

BACKGROUND LITERATURE

Chapter 1

BODY CENTRE OF MASS AS A USEFUL GAIT ANALYSIS BODY FEATURE

1. BIOMECHANICS: GENERAL CONCEPTS AND GOALS

1.1. A brief history of Biomechanics

Although Biomechanics is a relatively young and dynamic field, its history can be traced back to the fifteenth century (Robertson et al., 2004; Hoffman, 2005; McGinnis, 2005). The most definitive step for further biomechanical advances occurred during the *Scientific Revolution* which culminated in the work *De Motu Animalium* by Giovanni Alfonso Borelli (1608-1679), the real founding father of Biomechanics (McKinon et al., 2004).

In 1836, the two brothers Eduard (1795-1881) and Wilhelm (1804-1891) Weber published their treatise *Die Mechanik der menschlichen Gehwerzeuge* (*On the Mechanics of the Human Gait tools*) containing almost 150 hypotheses about human gait (Scheleihauf, 2004). In the last decades of the 19th century, Etienne Jules Marey (1838-1904) transformed the study of locomotion from an observational science to one based on quantification (Richards, 2008). Edward Muybridge (1830-1904) began a lifetime devotion to documenting the sequential motion of humans and animals (Muybridge, 1899). In 1891, Wilhelm Braune and Otto Fischer made precise mathematical analysis possible by conducting the first three-dimensional analysis of the human gait (Braune et al., 1987; Mündermann et al., 2006).

In addition, researchers interested in the biomechanics of human movement were active throughout the 20th century. Nicholas Bernstein (1896-1966) developed a method for measuring movement based on mathematical analysis. Archibald Hill (1886-1977) conducted studies into the mechanics of sprinting in the 1920s; this work was continued by Wallace Fenn (1862-1932) in the 1930s. Thomas Cureton (1901-1992) also wrote about the mechanics of swimming (1930) and various track and field skills in the 1930s. Bresler and Frankel (1950) were the first to carry out a mechanical analysis of walking (Richards, 2008).

1.2. What is Biomechanics?

1.2.1. The definition

The word Biomechanics can be divided into two parts: the prefix *bio-* and the root word *mechanics* (Hoffman, 2005). The prefix *bio* indicates that Biomechanics has something to do with living or biological systems. The root word *mechanics* indicates its relationship with the analysis of forces and their effects (see par. 1.2.2 below). In other words, Biomechanics is ‘the application of mechanical principles on living organisms’ (Lippert, 2006).

As shown, Biomechanics can be considered an *inter-discipline* which involves the basic body of knowledge ranging from Physics, Chemistry, Mathematics, Physiology to Anatomy (Winter, 2005). A wide variety of physical movements are involved: from the gait in a healthy person (Winter, 1983; Collett et al., 2007) or those physically handicapped one (Hirokazu et al., 1987; Baker, 2006) to the performance of an elite athlete (McGinnis, 2005).

1.2.2. Mechanics *versus* Biomechanics

Mechanics can be divided into several branches: rigid-body mechanics, deformable-body mechanics, fluid mechanics, relativistic mechanics and quantum mechanics.

The objects concerned in this research project are human locomotion and its peculiarities. Thus, it is necessary to focus on only **rigid-body mechanics**. In fact, it is best suited for describing and explaining the gross movements of humans and implements in daily life activities (ADL), normal and pathological sport (Racic et al., 2009). In rigid-body mechanics, the objects investigated are assumed to be perfectly rigid; which is why, they do not deform by bending, stretching or compressing. Therefore, in describing and clarifying the gross movements in the human body, it will be necessary to consider the segments of the human body as rigid bodies that are linked together at joints (Winter, 1979). In reality, the segments of the body deform under the action of forces (Enoka, 2002). However, these deformations are usually minor and do not appreciably affect the gross movements of the limbs or the body itself so that it is possible to get away with considering the body as a system of linked rigid bodies (Enoka, 1994).

In *Biomechanics*, rigid-body mechanics could indeed be subdivided into: 1) *statics*: the mechanics of objects at rest or moving at a constant speed; and 2) *dynamics*: the mechanics of objects in speeding up motion. It is further subdivided into *kinematics* and *kinetics* (Winter, 1990a; Novacheck, 1995; McGinnis, 2005; see also par. 2.3 below). All basic kinematic and kinetic concepts will be covered in detail in two-dimensional analyses. In three-dimensional analyses, another vector direction has to be added so that three planes are developed (as suggested in Winter, 1979: coronal plane, sagittal plane and transverse plane).

Other variables that are used in the description and analysis of movement are:

- *electromyography*. The EMG is the primary signal to describe the input to the muscular system (Richards, 2008). It provides information regarding which muscles are responsible for a muscle moment or whether antagonistic activity is taking place (Cavagna, 1988; Zajac et al., 2003; Robertson et al., 2004; Nigg et al., 2005);

- *anthropometry* (see also chapter 6, par. 2.1). It is impossible to evolve a biomechanical model without data regarding masses of limb segments, location of centre of masses, segment lengths, centres of rotation, moments of inertia, radius of gyration and so on (Winter, 2005; Lippert, 2006; Racic et al., 2009).

2. GAIT ANALYSIS: AN OPEN QUESTION?

2.1. Introduction

Locomotion is a complex act arising from the coordination of multiple mechanisms and couplings of the neuromuscular system, including the motor cortex, cerebellum, basal ganglia and feedback from vestibular, visual and peripheral receptors (Dierick et al., 2004; Jordan et al., 2006; Locomotion in Biomechanics - Wikipedia, the free encyclopedia, 2009). In humans, locomotor development it must be flexible enough to accommodate changing environmental demands and task constraints (Reisman et al., 2007).

Human locomotion is a phenomenon which constitutes the translation of the centre of gravity through space along a pathway requiring the least expenditure of energy supplies (Saunders et al., 1953; see par. 2.4 below).

Gait analysis is the study of animal locomotion, including human locomotion (Davis et al., 1991; Perry, 1992; Craik et al., 1995; Hreljac, 1995a; Whittle, 2002; Baker, 2006; Hreljac et al., 2007; Racic et al., 2009). It could be commonly used to help athletes run more efficiently and to identify posture-related or movement-related problems in people with injuries (Perry, 1990). The study of gait analysis encompasses quantification (e.g. introduction and analysis of measurable parameters of gaits), as well as interpretation (e.g. drawing various conclusions about the animal - health, age, size, weight, speed and so on -) from its gait.

As regards this research project, we were particularly interested in **human walking** and **running**. In fact, our research focuses highly on these two different types of locomotion so the following sections are entirely related to these gaits (see par. 2.4 below onwards).

2.2. A brief history of gait analysis

With the development of photography, it became possible to capture image sequences which revealed details both of animal and human locomotion. These movements could not be noticed by watching the movement with the naked eye (Winter, 2005; Racic et al., 2009). E. Muybridge was one of the first pioneers of this in the early 1900s (see also chapter 4, par. 3).

Although much early research was done using cameras, the widespread application of gait analysis to humans with pathological conditions such as cerebral palsy (Gage, 1991; Massaad et al., 2004; Bennett et al., 2005; McNee et al., 2006), Parkinson's disease, amputees (Detrembleur et al., 2005) and neuromuscular disorders (Gutierrez et al., 2003), began in the 1970s with the availability of video camera systems which produce detailed studies within realistic cost and time constraints.

The development of treatment regimes, often involving orthopedic surgery, based on gait analysis results, advanced significantly in the 1980s (Baker, 2006). As expected, although the technological achievements have spread the use of gait analysis in both sport and clinics, it is important to note that these approaches still suffer from intrinsic technical complication and lack of clear understanding of the significance of the data by the general user.

As a result, in spite of the quantitative data that can be obtained from biomechanical approaches, it is the experienced eye that plays the most relevant role in prognostic/diagnostic processes in the clinical context or in sports (Minetti, 2006).

2.3. Equipment and techniques

Gait analysis commonly involves the measurement of the movement of the body in space (kinematics) and the forces involved in producing these movements (kinetics).

Kinematics can be recorded using a variety of systems and methodologies (Sutherland et al., 1994): photography, video recordings, passive and active marker systems (see also chapter 4, par. 3.2 and 3.3). Furthermore, to calculate movement *kinetics*, it is possible to use floor load transducers, also known as force plates, which measure the ground reaction force, including both magnitude and direction (Winter, 2005), the forces exerted by each muscle group and the net moment around each joint at every stage of the gait cycle (see par. 2.5 below). The computational method used for this is referred to *inverse dynamics* (Vaughan, 1999; Racic et al., 2009). To detect the activity and contribution of individual muscles to movement, it is necessary to investigate the electrical activity of muscles (Forssberg et al., 1994).

Gait analysis deviations from normal kinematic, kinetic or EMG patterns are used to diagnose specific conditions and predict the outcome of treatment.

2.4. The main types of human locomotion: walking and running

Walking and running are prominent modes of locomotion for terrestrial vertebrates (Snyder et al., 2008). Therefore, the following sections will focus on both walking (par. 2.4.1) and running (par. 2.4.2).

2.4.1. Walking

Human walking is a complex task that requires coordination, flexibility and adaptability of a number of different muscles acting on different joints (Hausdorff et al., 1995; Mahaudens et al., 2009; O'Connor et al., 2009).

From a biomechanical point of view, walking is the most common gait, where the two feet are on the ground at any given time (Alexander, 1992; Winter, 1992; Sutherland et al., 1994; Koopman et al., 1995; Prince et al., 1997; Bastien et al., 2003; Lippert, 2006; Paroczai et al., 2006; Segers et al., 2006). In fact, one characteristic phase of walking (see par. 2.5 below) is the double support interval. This interval decreases as the speed of the subjects increases until it disappears; the subject is then considered to be running (Messenger et al., 1994).

As a consequence, normal walking depends on a continual interchange between mobility and stability (Hausdorff et al., 1995; Holt et al., 1995). Moreover, translation of the body centre of mass from one place to another is a fundamental objective of walking (Eames et al., 1999). In fact, normal human walking is characterized by a periodic vertical displacement of the BCOM (by 4 cm at every step) that moves through a complete cycle of vertical motion with each step during each stride (Detrembleur et al., 2000; Gard et al., 2001; 2004; see also par. 3.6.2).

Learning to walk occurs during the first year of life and reaches maturity around 7 until 60 years. Very young children and healthy adults walk quite differently to healthy young adults (Sutherland et al., 1980; Jeng et al., 1997; Kang et al., 2009). In detail, the development of mature walking is characterized by changes in the duration of single-leg stance, walking speed, cadence, step length, step width and the ratio of pelvis span to ankle spread (Grieve et al., 1966; Buchner et al., 1996; Hausdorff et al., 1999). The duration of single-leg stance, step length and walking speed is lower in very young children (below the age of 7) than in young adults (Dierick et al., 2004; Lippert, 2006). Cadence and step width are both higher.

Elderly walking performance then starts to decline and gradually slows down (Murray et al., 1969; Hageman et al., 1986; Buchner et al., 1996; Cromwell et al., 2002; Kang et al., 2009). Step length, walking speed, the duration of single-leg stance and muscle performance all decrease with advancing age (Bastien et al., 2003), while step width increases to widen the base of support (Himann et al., 1988; Sutherland et al., 1994; Buchner et al., 1996; Judge et al., 1996a; 1996b; Prince et al., 1997; Kerrigan et al., 1998; Hausdorff, 2004).

2.4.2. Running

Running is considered to occur when, at some points in the stride, all feet are off the ground in a moment of suspension (Novacheck, 1998; Paroczai et al., 2006; Segers et al., 2006; Steudel-

Numbers et al., 2007). Technically, moments of suspension occur in both running gaits (such as *trot*) and leaping gaits (such as *canter* and *gallop*). As previously discussed in walking, elderly running performance starts to decline and gradually slows down, as well (Korhonen et al., 2009).

The running velocities influences the stride frequency (Mero et al., 1992). In running at a speed of about 3 m/s, the amplitude of the vertical oscillations of the BCOM ceases to be symmetric (Saibene et al., 2003); this is a critical speed, independently of age.

2.4.3. Walking *versus* running

When walking faster and faster, humans will spontaneously start running. Generally, both gaits are distinguished from each other on the basis of the difference in dynamics of the body's centre of mass (Starke et al., 2009). To be specific, the demarcation between walking and running occurs when periods of double support during the stance phase of the gait cycle give way to two periods of double float at the beginning and at the end of the swing phase of gait (Hreljac, 1993a; Segers et al., 2006; see par. 2.5 below). Generally as speed increases further, initial contact changes from the heel to the toe (Diedrich et al., 1995; Lippert, 2006).

In conclusion, the transition from walking to running occurs when the double support period(s) is eliminated (Alexander, 1992; Brisswalter et al., 1996b; Alexander, 2003; Hreljac et al., 2007; Segers et al., 2007a; Steudel-Numbers et al., 2009). It has been demonstrated that walking is optimal (minimum metabolic energy per unit distance) at a given speed (about 1.4 m/s for level walking; Thorstensson et al., 1987; Gordon et al., 2009). Some of the mechanical determinants have already been identified in previous literature, as far as level walking is concerned (Cavagna et al., 1976; Cunningham et al., 1982; Saibene et al., 2003).

The pendulum-like mechanism describing the body centre of mass vaults was shown to be best exploited by a speed range close to the optimum speed (see par. 4.3 below).

One common movement pattern transition that interests some motor control researchers is the transition in human gait between walking and running (Ardigò, 1992; Brisswalter et al., 1996b) when the speed of travel exceeds a threshold of some 2 m/s (≈ 7 km/h; Seay et al., 2006). This transition occurs apparently quite automatically (without any conscious awareness), but the question of interest is what triggers the neuromuscular system to make this pattern reorganization (Hanna et al., 2000). As a result, speed (or perhaps more strictly, frequency) is generally regarded as a control parameter for human gait transition (Raynor et al., 2002).

Classical studies of walking, trotting and galloping in horses by the biologist Hoyt and Taylor in the early 1980s (Minetti et al., 1994) suggested that physiological efficiency was the 'key controller', with the animals favoring speeds of travel that were optimal for minimizing required energy expenditure. In humans, the minimization of energetic costs is one of a number of potential control variables (Scheleihauf, 2004).

The main features of human gait that vary with speed, in highly predictable ways, include: a) stride length, the distance between corresponding points on successive footprints of the same foot (Alexander, 1992; Segers et al., 2006; see also chapter 10, par. 3); b) duty factor, the fraction of the duration of the stride period in which each foot is on the ground (Alexander, 1989; Segers et al., 2007b; Starke et al., 2009; see also chapter 10, par. 5); and c) shape factor, a ratio of Fourier coefficients that describes the time course of the force exerted on the ground (Minetti et al., 1997).

2.4.4. Factors affecting human locomotion

Human locomotion (both walking and running) is affected by various and different factors: a) speed of locomotion and consequent speed transition from one form of locomotion to another (Cavagna et al., 1976; 1981; 1983; Minetti et al., 1994; Schepens et al., 1998; Minetti et al., 2003; Seay et al., 2006); b) gradient: level, uphill and downhill (Grasso et al., 1998; Minetti et al., 2001; 2002); c) subject age (Cavagna et al., 1983; Crowe et al., 1996; Schepens et al., 1998; 2001); d) subject gender (males and females; Crowe et al., 1996); e) subject size (height and weight; Crowe et al., 1996); f) ground type (e.g. over-ground or treadmill locomotion; see also chapter 19, par. 1 and 2); g) eventual added-weight (Crowe et al., 1996; Saibene et al., 2003); h) the influence of gravity; and i) outdoor and indoor climatic conditions (Crowe et al., 1996).

Our experiments take some of these factors into consideration (see also chapters 3 and 5).

2.5. The gait cycle

2.5.1. Introduction

The gait (or stride) cycle (GC) is ‘the interval between sequential initial floor contacts by the same limb’ (Hansen et al., 2002; Whittle, 2002; Richards, 2008; ESMAC Hand Notes, 2009), whereas a step is recognized as the interval between sequential floor contacts by ipsilateral and contralateral limbs (Perry, 1989; 1992; Lippert, 2006; Racic et al., 2009).

Two steps make up each GC, which is *roughly symmetric* in normal individuals (Whittle, 2002); it is simply comprised of stance and swing phases (Figure 1.3; see par. 2.5.2 below).

Spatial and temporal parameters are illustrated in Figure 1.1a and 1.1b, respectively.

The spatial parameters of foot contact during gait should be considered (Du Chatinier et al., 1970; Zatsiorsky et al., 1994; Sekiya et al., 1997; Danion et al., 2003; Hausdorff et al., 2004; Reisman et al., 2007; Hernandez et al., 2009; Hillman et al., 2009; Sibley et al., 2009): a) step length, the distance between two consecutive initial contacts by different feet; b) stride length, the distance between two consecutive initial contacts by the same foot; c) foot angle, the angle of foot orientation away from the line of progression; and d) base (or step) width, the medial-lateral distance between the centre of each heel during gait. Two other parameters may easily be calculated using this information: e) cadence, the number of steps per minute (Webb et al., 2007; Houdijk et al., 2009) and f) average velocity (Alexander, 2003).

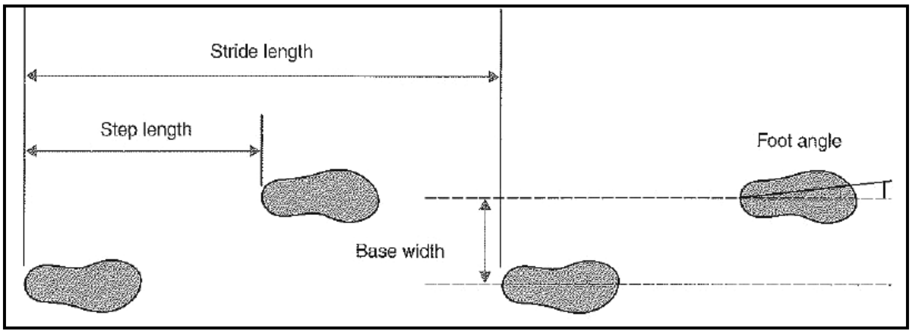


Figure 1.1a. Gait analysis spatial terminology, in Richards (2008).

Foot contact times are important temporal parameters (Hogberg, 1952; Maruyama et al., 1992). Step time and stride time are defined as the time between two consecutive initial contacts by different legs and the time between two consecutive initial contacts by the same leg, respectively. Moreover, single and double support time may be defined as the time when one foot and both feet are in contact with the ground, respectively (ESMAC Hand Notes, 2009).

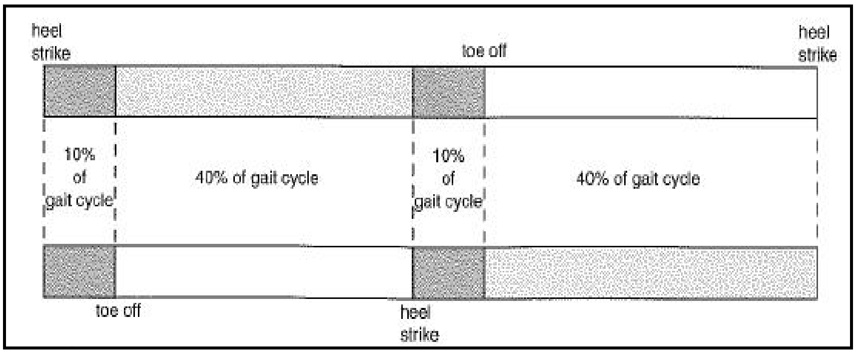


Figure 1.1b. Gait analysis temporal terminology, in Richards (2008).

The gait cycle is the basic unit of measurement in gait analysis.

2.5.2. Single phases of the gait cycle

There are two main phases of the gait cycle: stance phase and swing phase (Perry, 1992; Hansen et al., 2002; Kuo, 2007; Richards, 2008; Starke et al., 2009).

The **stance** phase is the activity that occurs when the foot is in contact with the ground: it begins with heel strike of one foot and ends when the foot leaves the ground (ESMAC Hand Notes, 2009).

To be specific, it is subdivided into 3 segments, including initial double stance (corresponding to loading response), single limb stance (corresponding to mid stance and terminal stance) and terminal double limb stance (corresponding to pre-swing). Each double stance period accounts for 10% of the GC, while single stance normally represents 40% (60% total).

The **swing** phase for this same limb is the remaining 40% of the GC. It occurs when the foot is not in contact with the ground: it begins as soon as the foot leaves the floor and ends when the heel of the same foot touches the floor (ESMAC Hand Notes, 2009).

Slight variations occur in the percentage of stance and swing related to gait velocity. The duration of each aspect of stance decreases as walking velocity increases (Rosenrot et al., 1980; Detrembleur et al., 2005; Racic et al., 2009; see also chapter 10, par. 5).

Vertical, anterior/posterior and medial/lateral ground reaction forces (Figure 1.2) can also give researches important information about the overall functioning of each phase and of the lower limb (ESMAC Hand Notes, 2009).

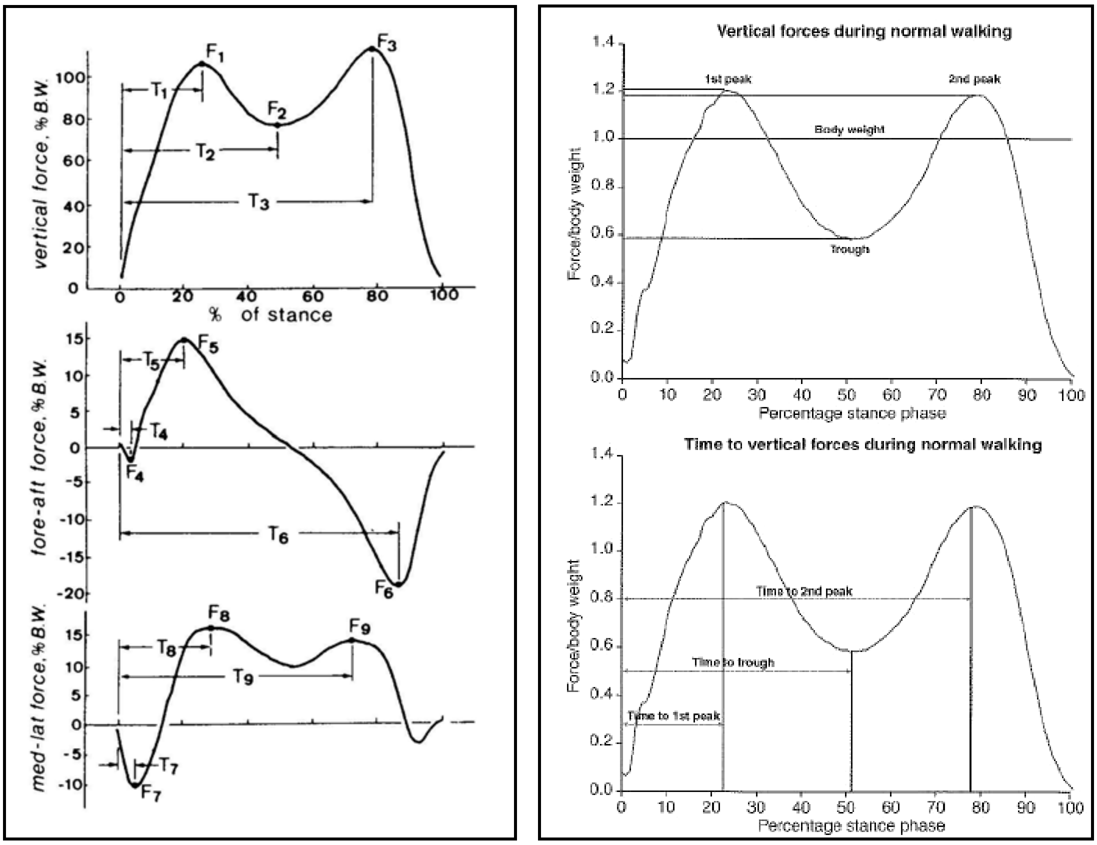


Figure 1.2. Ground reaction force components (on the left)
and a focus on vertical forces (on the right) during normal walking.

The gait cycle has two periods of double and single support. A period of non-support, that is, a time during which either foot is in contact with the ground, does not occur during walking; however, it does occur during running.

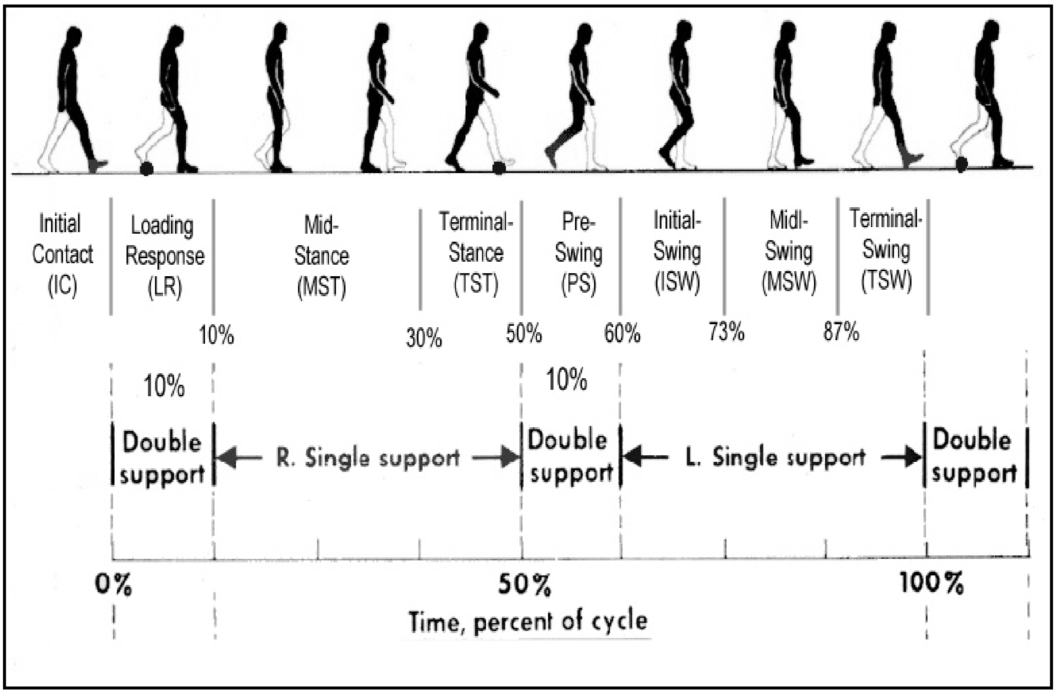


Figure 1.3. Phases of a gait cycle, in Perry (1989).

To be precise, a consistent sequence of motions is performed at each of the lower extremity joints during locomotion. Each stride contains 8 (Larsson et al., 1980) relevant phases (Figure 1.3).

1. The initial stance phase is made up of 5 gait phases, with the remaining 3 phases occurring during swing. The first 2 gait phases (0-10% GC: initial contact and loading response) occur during initial double support. The first half of single support is termed mid stance (10-30% GC): body passes over the stance leg. It is concerned with progression of the body centre of mass (see par. 3 below) over the support foot. This trend continues through terminal stance (30-50% GC), a phase which includes heel rise of the support foot (Kerrigan et al., 2000) and terminates with contra-lateral foot contact.

2. The final stance element, pre-swing (50-60% GC), begins with terminal double support and ends with toe-off of the ipsilateral limb (Racic et al., 2009).

3. Three unique phases characterize swing, including initial swing (60-73% GC), mid swing (73-87% GC) and terminal swing (87-100% GC) (ESMAC Hand Notes, 2009).

Because the stance phase in walking is longer than 50% of the gait cycle, there are two periods of double support when both feet are on the ground: one at the beginning and one at the end of the stance. In running, toe off occurs before 50% of the gait cycle is completed. There are no periods when both feet are in contact with the ground. Instead, both feet are airborne twice during the gait cycle, one at the beginning and one at the end of swing, referred to as double float. The timing of toe off depends on speed (Novacheck, 1998).

3. BODY CENTRE OF MASS (BCOM): A POSSIBLE SOLUTION

3.1. Centre of gravity *versus* centre of mass

The terms *centre of gravity* (COG or COP, *centre of pressure*) and *centre of mass* (COM) are often used interchangeably (Cavagna, 1988; Eng et al., 1993; Hasan et al., 1996; Winter et al., 2003; Gage et al., 2004; Robertson et al., 2004; Hof, 2008). Thus, first of all, it is necessary to distinguish between these two similar but very different physical concepts (Figure 1.4).

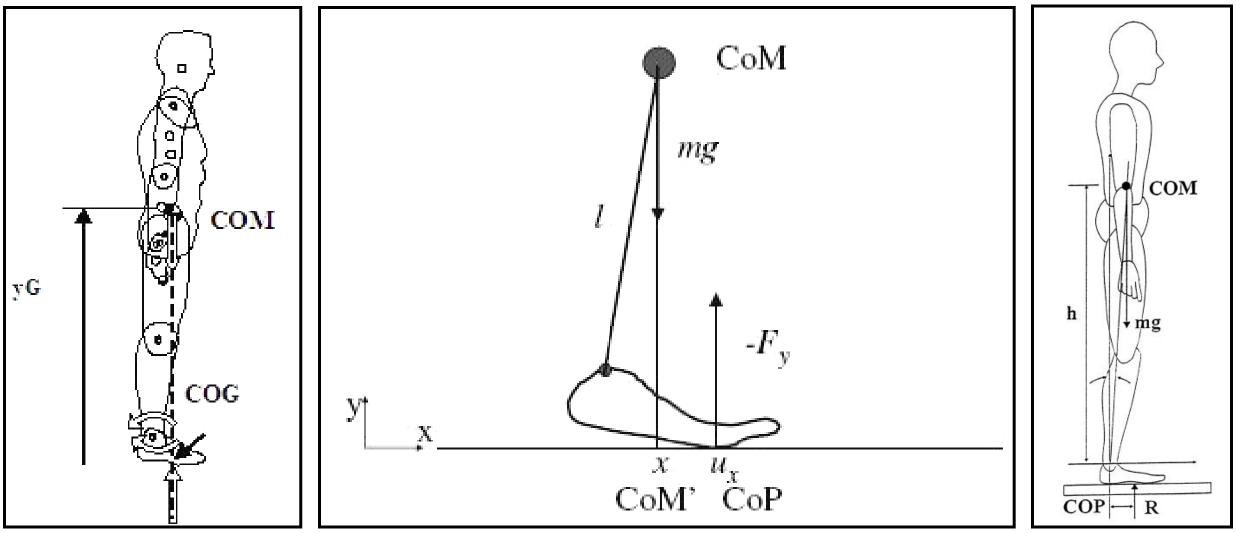


Figure 1.4. Centre of gravity (COG) or centre of pressure (COP) *versus* centre of mass (COM).

The *centre of gravity* is the point in a body or system around which its mass or weight is evenly distributed or balanced and through which the force of gravity acts (McGinnis, 2005; Winter, 2005). Thus, it is an imaginary point in space, not a physical entity. The vertical line through the centre of gravity is called the *gravity line*.

On the other hand, the *centre of mass* is the imaginary point in a body or system in which its mass is located (Winter, 1979; Richards, 2008). In the case of the human body, the centre of gravity (COG) coincides with the centre of mass (COM; Zatsiorsky, 2002).

This research project will focus on the kinematic variables of the centre of mass of the human body (BCOM) over time and space domains.

3.2. Centre of mass definition

In 1679 Giovanni Alfonso Borelli determined the position of the BCOM along the longitudinal axis by placing naked subjects supine on a platform which was balanced on a narrow ridge: one of the first experiments in human biomechanics.

Since then, some researchers have determined the BCOM position.

In Physics, the *centre of mass* of a system of particles is a specific point at which, for many purposes, the system mass behaves as if its masses are concentrated. Thus, the three-dimensional trajectory of the centre of mass is a function only of:

1. the three-dimensional positions of the particles that constitute the system;
2. the masses of these particles.

These two physical variables are necessary to isolate, locate, define and calculate the exact position of the BCOM.

3.3. Centre of mass calculation

3.3.1. Introduction: mass and density

Mass (*m*) is the fundamental property of matter to resist a change in velocity (Zatsiorsky, 2002; McGinnis, 2005); see Equation [1.1] below). The SI unit of mass is the kilogram (kg). Density (ρ) is the amount of mass per unit of volume V (Winter, 2005; Richards, 2008; see Equation [1.2] below).

If the masses contained in any two equal volumes of a given body are equal, the body is said to be a *homogeneous* body; otherwise, it is a *heterogeneous* body. Since bones, muscles, internal organs and fat tissues have different densities, the human body is heterogeneous (Zatsiorsky, 2002).

3.3.2. How it is possible to calculate the centre of mass

As the total body mass increases, so does the mass of each individual segment. Therefore, it is possible to express the mass of each segment as a percentage of the total body mass (see also chapter 6, par. 2.1). The total mass M of the segment is

$$M = \sum_{i=1}^{n} m_i \quad \text{[Eq. 1.1]}$$

where m_i is the mass of the ith section. Moreover, m_i could also be expressed as:

$$m_i = \rho_i \bullet V_i \quad \text{[Eq. 1.2]}$$

where ρ_i is just the density of the ith section; and V_i is the volume of the ith section. If the density ρ at every point in a body is assumed to be uniform, the total mass M of the body can be calculated as the integral over the body segments:

$$M = \int (\rho \bullet dV) \quad \text{[Eq. 1.3]}$$

The centre of mass is such that it must create the same net gravitational moment of force at any point along the segment axis as did the original distributed mass (Winter, 2005). The product of a mass m_i concentrated in a point P_i by the distance x_i from a given plane *yz* is:

$$M_i = m_i \bullet x_i \quad \text{[Eq. 1.4]}$$

This product is known as *the moment of the first order* of a material particle with respect to *yz* plane (Zatsiorsky, 2002). It is now possible to represent the complex distributed mass by a single mass *M* located at a distance *x* from one end (proximal or distal) of the segment (Winter, 2005). For a continuous mass, it is possible to omit the subscript *i* and replace mass *m* with density *ρ*, as the relationship presented in the Equation [1.2]. The integral

$$M = \int (m \bullet x \bullet dV) = \int (\rho \bullet x \bullet dV) \quad \text{[Eq. 1.5]}$$

is known as *the mass moment of the first order* of the body with respect to the *yz* plane (Zatsiorsky, 2002). There always exists a point with respect to which the mass of the first order in any plane equals zero. When the mass moment of the body is computed with respect to this point, the integral of the positive *mx* products in the Equation [1.5] equals the integral of the negative products, and they cancel each other. This is the point called the centre of mass of the body (BCOM). Therefore, the Cartesian coordinate of the BCOM is obtained by the Equation:

$$r_{BCOM} = \frac{1}{M} \bullet \int (\rho \bullet r \bullet dV) \quad \text{[Eq. 1.6]}$$

where r_{BCOM} is the three-dimensional position vector of the BCOM. If the body is homogeneous, the COM coincides with its centre of volume, the *centroid*.

Finally, when the masses (m_i) and the location of the BCOM of the individual body parts along a given coordinate (r_i) are known, the location of the total BCOM can be easily found from:

$$r_{BCOM} = \frac{\Sigma m_i \bullet r_i}{\Sigma m_i} \quad \text{[Eq. 1.7]}$$

3.3.3. Centre of mass of a multi-segmented system

With each body segment in motion, the centre of mass of the total body is continuously changing with time. It is therefore necessary to recalculate it after each interval of time, and this requires a good knowledge of the trajectories of the centre of mass of each body segment.

A two-segment system where the centre of mass is located in a particular point in time is indicated in Figure 1.5 (Centre of Mass in Physics - Mathwords, the mathematical encyclopedia, 2009). This is a simplified situation, in which only one dimension (x) is considered:

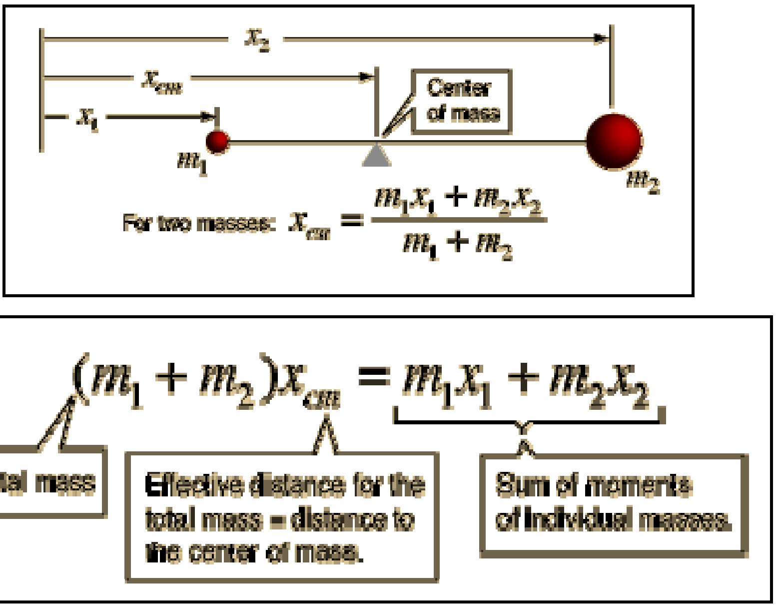


Figure 1.5. Centre of mass for a system of two masses, moving in only one dimension.

Anyway, this simple example can be extended to other more complex and various conditions, in which the other two dimensions (y and z) are considered.

Therefore, the term r_{BCOM} is determined by the combination of these different equations:

$$x_{BCOM} = \frac{\Sigma m_i \bullet x_i}{M} \text{ [Eq. 1.8a] or } x_{BCOM} = \frac{1}{M} \bullet \int (\rho \bullet x \bullet dV) \text{ [Eq. 1.8b]}$$

$$\text{and } x = \frac{1}{n}(\cos\varphi_1 + \cos\varphi_2 + \ldots + \cos\varphi_n) \text{ (rectangular coordinate) [Eq. 1.8c]}$$

$$y_{BCOM} = \frac{\Sigma m_i \bullet y_i}{M} \text{ [Eq. 1.9a] or } y_{BCOM} = \frac{1}{M} \bullet \int (\rho \bullet y \bullet dV) \text{ [Eq. 1.9b]}$$

$$\text{and } y = \frac{1}{n}(\text{sen}\varphi_1 + \text{sen}\varphi_2 + \ldots + \text{sen}\varphi_n) \text{ (rectangular coordinate) [Eq. 1.9c]}$$

$$z_{BCOM} = \frac{\Sigma m_i \bullet z_i}{M} \text{ [Eq. 1.10a] or } z_{BCOM} = \frac{1}{M} \bullet \int (\rho \bullet z \bullet dV) \text{ [Eq. 1.10b]}$$

In this way, it becomes possible to define and calculate the COM of any multi-segmented system (Figure 1.6). We were interested in the centre of mass of the human body, in particular. Indeed, estimation of a COM in multi-segmented human body requires kinematic measurement of all body segments (proximal and distal) displacements and an anthropometric model of the body (Winter, 2005).

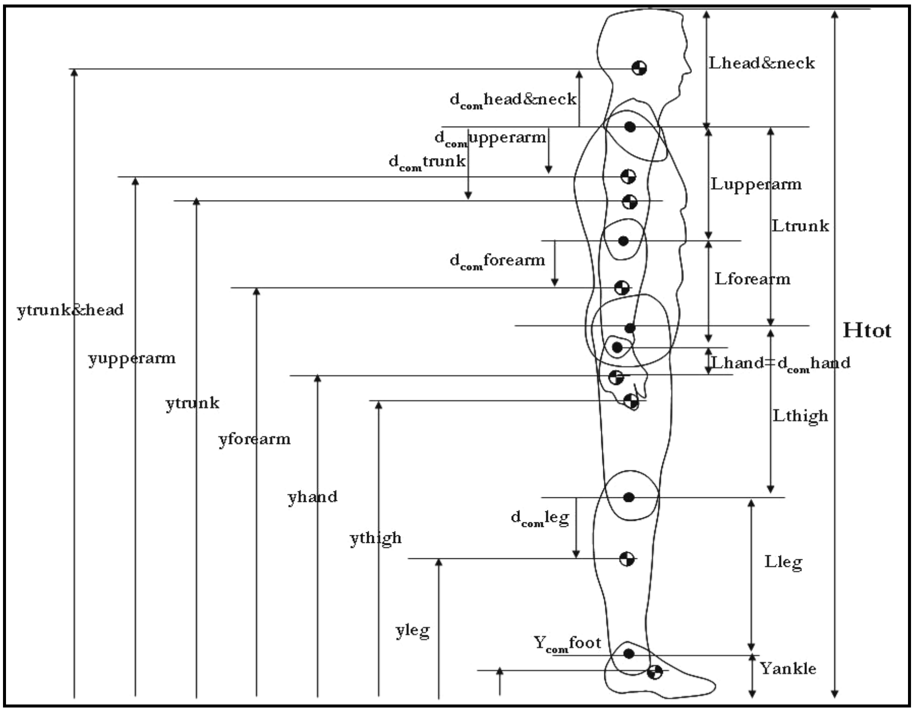


Figure 1.6. Centre of mass of each segment body during standing posture.
Clearly, only two dimensions (x and y) are considered.

3.4. Different locations of the centre of mass

3.4.1. Relationship with the geometry of the system

The COM position is related to the geometry (in terms of masses and three-dimensional positions of these masses) of the object or body:

a) in the case of a rigid body, the position of its centre of mass is fixed in relation to the object (but not necessarily in contact with it);

b) in the case of a loose distribution of masses in free space, the position of the centre of mass is a point in space among them that may not correspond to the position of any individual mass;

c) in the context of an entirely uniform gravitational field, the centre of mass is often called the *centre of gravity*, the point where gravity can be said to act (see par. 3.1 above);

d) in the case of the human body, the centre of mass (BCOM) does not always coincide with its intuitive geometric centre, and one can exploit this freedom (i.e. when bending over a bar in a high jump; Zatsiorsky, 2002; McGinnis, 2005; Richards, 2008).

In general, a woman's centre of mass is slightly lower than a man's (i.e. from 0.5% to 2.0%), because women have larger pelvic girdles and narrower shoulders compared to men (Zatsiorsky, 2002; McGinnis, 2005). Thus, a woman's centre of mass is approximately 55% of her height from the ground, whereas a man's centre of mass is approximately 57% of his height.

Infants and young children have a greater centre of mass relative to their height because of their relatively larger heads and shorter legs.

3.4.2. Centre of mass and stability

The knowledge of the exact position of the COM of an object or a body is very important to establish the stability of the system (Winter et al., 1990b; 1990c; Winter 1995a; 1995b; Pai et al., 1997; Stapley et al. 1998; Alexander, 2003; Hof et al., 2005; O'Connor et al., 2009).

But *how can we define the stability of an object?* Stability is 'the capacity of an object to return to equilibrium or to its original position after being displaced' (McGinnis, 2005).

And moreover, *what factors affect stability?* The stability of an object is affected by 'the height of the centre of mass, the size of the base of support and the weight of the object' (Richards, 2008).

Finally, it becomes very important to remember that the most stable stance or position a person can assume is that in which the body centre of mass is located.

3.5. Literature focusing on the body centre of mass

3.5.1. Introduction

In this section, we will deal with details on different methods estimating the three-dimensional trajectory of the BCOM. In fact, the literature presents and highlights different methodologies to estimate and quantify the BCOM.

Some studies have proposed using kinetics instrumentation (like force platforms) to study and define the three-dimensional trajectory of the BCOM (see par. 3.5.3 below; Craik et al., 1995; Kingma et al., 1995; Dierick et al., 2004; Gard et al., 2004; Hallemans et al., 2004; Lafond et al., 2004; Zok et al., 2004; Mian et al., 2006a; Feng et al., 2009). Others have suggested employing kinematics instrumentation (like a motion capture system) to estimate the position of the BCOM (see par. 3.5.4 below; Thirunarayan et al., 1996; Richards, 1999; Vaughan et al., 1999; McKinon et al., 2004; Gutierrez-Farewik et al., 2006; Gordon et al., 2009; Gullstrand et al., 2009).

As a result, *what are the main techniques to characterize the three-dimensional position of the BCOM? Does a unique method exist*? Furthermore, *what is the best method to estimate the position of the BCOM*? It becomes very difficult to try and give a unique and single answer to these crucial questions. In fact, in the literature, different schools of thought coexist. However, to better understand how this important biomechanical variable could be investigated, this has to be clarified.

In this section, therefore, we will submit a brief review of the literature concerning these different methods. As will be seen, it is not easy to define the best and the most validated technique.

3.5.2. What is the best method to estimate the position of the body centre of mass?

Over years, different researchers have tried to satisfactorily reply to this crucial question.

Firstly, the agreement between 1) the segmental method and 2) the more direct reaction-board measurement method has been assessed (McKinon et al., 1994). It has been demonstrated that ‘the segmental method agrees with a more direct technique of known accuracy, the reaction-board method, most closely when measuring averaged oscillation over repeated strides’.

In other important studies, different methods for estimating vertical displacement of the BCOM have been compared: a) the force plate, the sacral marker and the segmental analysis methods (Thirunarayan et al., 1996). In the end, it has been found that: 1) the force plate is an insensitive and inaccurate method; while 2) the sacral marker and the segmental analysis methods are the simplest, the most inexpensive, most effective as well as most accurate tools; b) the force plate, the sacral marker, the segmental analysis and the pelvis methods, as well (Saini et al., 1998). The sacral marker method, which is the most simple, can provide essentially the same estimate as the more complicated reconstructed pelvis and segmental analysis methods. However, the force plate provided information that is statistically different form the results of the kinematic methods: data had a lower range and a different distribution; c) more recently, the sacral marker method, the body segmental analysis technique and the double integration of force platform data (Gard et al., 2004). The results showed that at the lowest walking speed the vertical excursions calculated by all three techniques were similar, but at the faster speeds the sacral marker significantly overestimated compared with the other two methods; d) the kinematic method, the zero-point-to-zero-point double integration technique and the centre of pressure low-pass filter method (Lafond et al., 2004). The results show that: 1) the zero-point-to-zero-point double integration method is comparable to the kinematic method; and 2) this technique is attractive from the clinical perspective because it requires only one force plate to determine the centre of pressure-centre of mass variable, which has been demonstrated to be highly reliable. As regards the double integration technique, it has been properly used to estimate whole body centre of mass displacement from signals of a single force platform (Zok et al., 2004). This technique’s COM displacement estimates were more repeatable and up to 50% more accurate than those of a regular and time-reversed double integration; and e) recently, it has been demonstrated that the position transducer and the motion capture system show strong correlation, whereas the accelerometers show a variability increasing with velocity (Gullstrand et al., 2009). Finally, the three-dimensional BCOM accelerations have been recorded by means of a tri-axial accelerometer and gyroscope to obtain BCOM velocities, displacements and fluctuations in potential and kinetic energies (Peyrot et al., 2009).

3.5.3. Kinetics data describing the movement of the body centre of mass

As previously shown (see par. 3.5.1 above), some researches have proposed to use kinetics instrumentation to study the three-dimensional trajectory of the BCOM. Regarding this technique, many studies have been proposed in literature. In order to better explain the importance of this technique, the main investigations will be briefly presented and discussed.

Firstly, the kinetics of the BCOM has been widely analysed during level walking in normal and pathological gaits (Iida et al., 1987). Importantly, the parameters obtained from the displacements and the energy variations of the BCOM are considered useful in the evaluation of stability and efficiency for pathological gaits.

In another study, two force plates were used to determine the cyclic oscillation of the BCOM in young female adults walking at their preferred speed (Crowe et al., 1995). Positively, it has been found that: a) good approximations to the oscillations may be obtained from formulae containing just the first- and second- order Fourier coefficients of the ground reaction forces taken over a complete walking cycle; and b) the amplitudes can be used as sufficient parameters to characterize the BCOM oscillations, because the symmetric components of the oscillations have consistent mutual phase relations for normal subjects. The scientific validity of these findings could help in understanding our method of analysis, as well (see chapter 6, par. 3 and 4).

Furthermore, two embedded force platforms were used to measure the ground reaction forces in investigating symmetry of weight distribution and avoidance of lateral displacement of the BCOM during standing-up in healthy subjects (Hesse et al., 1996). Importantly, the weight distribution between both lower limbs and the medial/lateral displacement of the BCOM before and after seat-off were significantly different across subjects thereby indicating a non-strict symmetrical motion pattern in healthy persons. In comparison, it should be noted that, in normal gait, the average lateral displacement of the BCOM amounts to ≈ 3 cm (see also par. 2.4.1 above).

Recently, the trajectory of the BCOM has been recorded by means of ground-reaction force in human level walking (Kokshenev, 2006). As expected, at speeds of 1.69 ± 0.70 m/s, the trajectory exhibits a crossover from the slow walk to the normal gait. The instability of the trajectory occurs at the maximal speed (3.33 ± 1.00 m/s) of the fast human walk, when contact with the ground ceases to exist.

Finally, the global kinematics and symmetry derived from the accelerometer measurements with those of BCOM movements were compared in healthy subjects walked barefoot over two different adjacent force-platforms at self-selected speeds (Meichtry et al., 2007). The results show that: a) BCOM anterior/posterior displacement lagged behind the trunk by 3.5% during the gait cycle; b) external power correlated highly between the trunk model and the centre of mass; and c) work and power asymmetry indexes correlated higher.

3.5.4. Kinematics data describing the movement of the body centre of mass

However, some researches have proposed to use kinematics instrumentation to study the three-dimensional trajectory of the BCOM. In order to better explain the importance of this other technique, the main investigations will be briefly presented and discussed.

Firstly, the 3D motion of the centre of the pelvis during walking, in normal adults of both sexes, has been recorded by a kinematic gait analysis system (Whittle, 1997). It has been showed how the centre of gravity moved in the opposite direction to the motion of the trunk. In addition, although as a first approximation the motion of the trunk and the motion of the BCOM are clearly similar, they are not exactly the same, since the BCOM is affected by the positions of the arms and legs, which move independently of the trunk.

Kinematic models of the body have been proposed to obtain the estimation of the total BCOM position, as well (Rabuffetti et al., 1999). Particularly, Thirunarayan's ('approximate model': 1996) and Zatsiorsky's ('analytical model': 1983) models have been compared. Results accounted for a better performance of the 'analytical' model with respect to the 'approximate' one, and the quantified accuracy of the latter model confirmed its accuracy for application in gait analysis.

By using kinematics technique, the effects of obstacle height on the motion of the whole BCOM and its interaction with the centre of pressure of the stance foot while negotiating obstacles have been investigated, too. Firstly, it has been demonstrated how stepping over the higher obstacles resulted in a) significantly greater ranges of motion of the BCOM in the anterior/posterior and vertical directions; b) a greater velocity of the BCOM in the vertical direction; and c) a greater anterior/posterior distance between the centre of mass and the centre of pressure (Chou et al., 2001). Secondly, by investigating whether elderly patients with imbalance (e.g. bilateral/unilateral vestibular weakness and unclear diagnosis) could be distinguished from healthy elderly subjects (Chou et al., 2003), it has been found: a) no significant group differences for the temporal-distance gait parameters during all testing conditions; and b) a significantly greater and faster lateral motion of the BCOM in elderly patients with balance disorders. More recently, to further investigate the impact of ageing on dynamic stability when negotiating stairs, both young men and healthy older men ascending and descending steps at a controlled cadence have been investigated (Mian et al., 2006a). Importantly, the maintenance of stability during gait is dependent on the ability to control the BCOM motion and, more specifically, its horizontal distance from the centre of pressure beneath the feet within appropriate limits, beyond which a corrective adjustment to gait would be necessary to avoid falling.

Finally, it has been widely stated that the kinematic method has some important advantages. For instance: a) it calculates accurately the BCOM position, not just displacement, both in globally and anatomically fixed reference frame; b) its accuracy does not depend on a steady state, symmetric gait; c) this technique makes it possible to measure how particular kinematics affect the BCOM movement; d) it is not depending on body mass estimation; and e) it can be also used to measure the BCOM during aerial activities (Gutierrez et al., 2003; Gutierrez-Farewik et al., 2006).

3.5.5. The body centre of mass displacement in normal gaits: some examples

The body centre of mass and its trajectory in space have been investigated in the context of:

a) normal walking (Craik et al., 1995; Della Croce et al., 2001; Meichtry et al., 2007). Importantly, the importance of every single phase and the possibility to consider the BCOM as a useful biomechanical indicator of normal walking has been verified;
b) sport-related issues (Cavagna et al., 1981);
c) walking in healthy children (Dierick et al., 2004; Holt et al., 2006). Particularly, vertical and lateral amplitudes of the BCOM when measured by leg length were greater for children under 4 years of age and forward amplitudes were greater for children under 7 years of age;
d) various pathological conditions (see par. 3.5.6 below).

3.5.6. The body centre of mass displacement in pathological gaits: some examples

The movement of the BCOM during walking can be an important descriptor of pathological gait and can be easily obtained during routine clinical gait analysis as a complementary measure to standard reporting (Gutierrez-Farewik et al., 2006).

As a result, the BCOM trajectories over space have also been investigated in:

a) children with cerebral palsy (Massaad et al., 2004; Bennett et al., 2005; Dezman et al., 2006; McNee et al., 2006; Toro et al., 2007). In all these studies, it has been demonstrated that: neither the topographical type, the severity of motor involvement nor the locomotory experience influenced BCOM displacements; however, vertical and forward BCOM displacements were significantly different between subjects without disabilities and children with spastic cerebral palsy (i.e. during the postural phase of gait initiation). In general, subjects with cerebral palsy have far less fine motor control of their BCOM than healthy peers and therefore complete simple balancing tasks less efficiently;

b) children with myelomeningocele (Gutierrez et al., 2003; Gutierrez-Farewik et al., 2006);

c) children with lumbosacral myelomeningocele (Eames et al., 1999). Positively, it has been shown how the use of a full body kinematic model gives a better representation of the BCOM than does a fixed point in the pelvis, especially in these pathological gait patterns;

d) children with spastic diplegia (Bennett et al., 2005). These subjects had a smaller energy recovery factor; a greater BCOM vertical excursion; a poorer phasic relationship between potential and kinetic energies, both of which contributed to greater mechanical work performed; and a less efficient gait;

e) patients with spinocerebellar ataxia (Bakker et al., 2006; Kung et al., 2009). They have shown greater peaks in BCOM anterior/posterior and lateral velocity: these peaks were correlated with increased trunk roll downhill and reduced uphill knee flexion velocity;

f) obese children and adolescents (Spyropoulos et al., 1991; Browning et al., 2007; Malatesta et al., 2009; Peyrot et al., 2009). All these studies have shown how the medial-lateral displacement of the BCOM is greater in obese subjects and significantly related to percent body fat;

g) adult amputees (Detrembleur et al., 2005). It has been demonstrated that the vertical displacement of the BCOM was greater in the trans-femoral group (i.e. depending on the absence of a knee flexion wave after the onset of weight loading) than in the trans-tibial. Moreover, the absence of a knee flexion-extension wave during the stance phase generated an increase in the vertical displacement without extra energy because of a conserved efficiency of the pendulum-like mechanism of walking.

3.6. Is it possible to use the body centre of mass to describe the gait cycle?

3.6.1. A general method of determining the trajectory of the body centre of mass

There are many points of the three-dimensional trajectory of the BCOM during human locomotion (walking and running, in particular) that could be obtained, as there are phases of movement whose co-ordinates have been established. These points are so close to each other that the whole trajectory of the BCOM can be clearly defined.

A general method of locating the BCOM for a precise phase of movement starts either from the centres of mass of the different segments of the body or from their principal points, as shown in Figure 1.4 above. Consequently, the total BCOM can be found by means of a geometrical construction, with respect to all movement directions and planes. It is important to remember that a human movement usually involves all three trajectories (anterior/posterior, vertical and medial/lateral) and planes (coronal, sagittal and transverse; as shown in Figure 1.7). In the following sections, the three-dimension positions of the BCOM will be deeply investigated and illustrated.

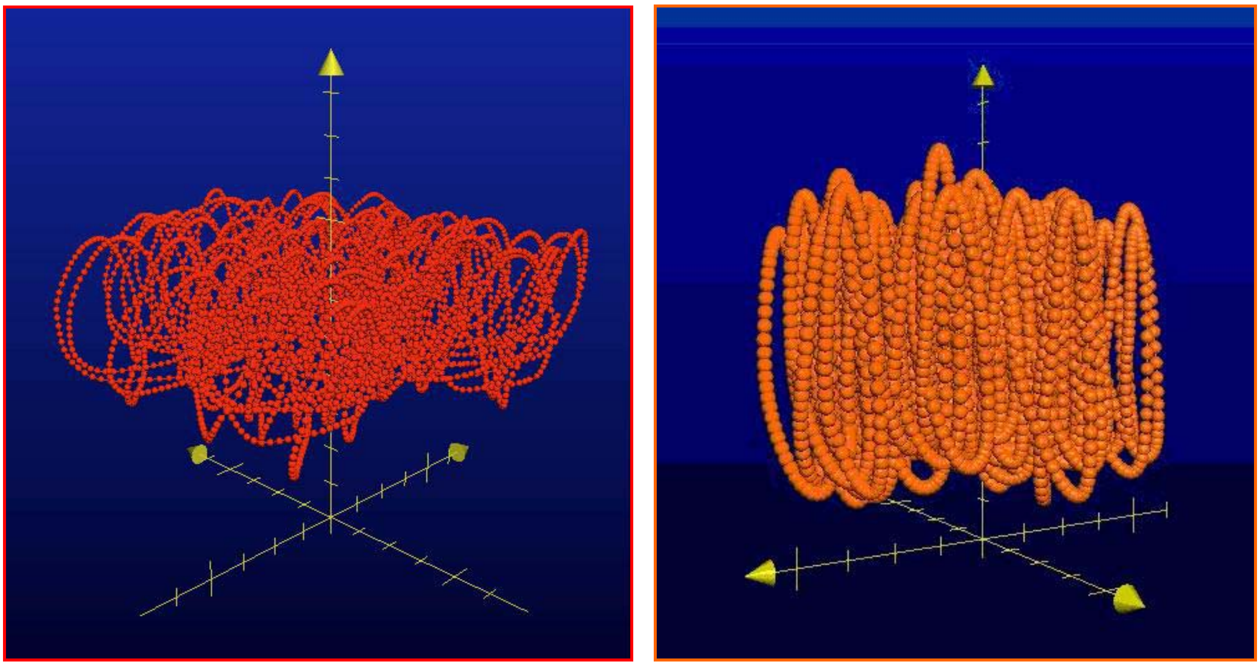
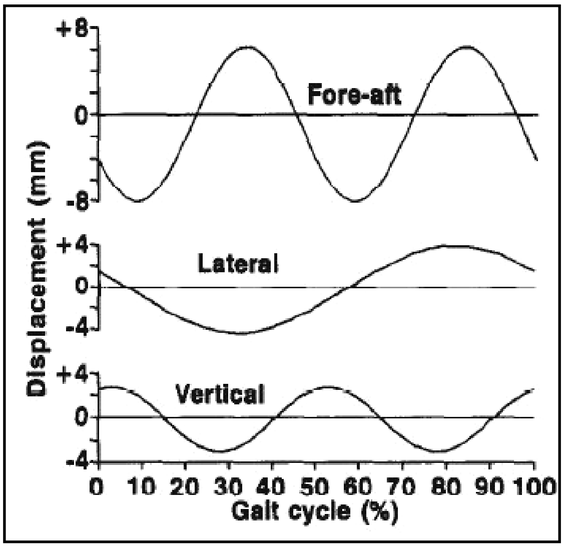


Figure 1.7. Examples of three-dimensional displacement of the BCOM independently of time variable.

3.6.2. The three-dimensional positions of the body centre of mass

The BCOM could be imagined travelling over the closed three-dimensional curve with its variable velocity, while this three-dimensional curve itself moves forward uniformly at the average velocity of gait. The movement of the BCOM is represented by three components: 1) one in the direction of the gait (Detrembleur et al., 2000; Mamoto et al., 2002; Orendurff et al., 2004); 2) one in the vertical direction of the gait (Belli et al., 1995; Kerrigan et al., 1995; Thirunarayan et al., 1996; Saini et al., 1998; Tesio et al., 1998; Gard et al., 2001; 2004; Ortega et al., 2005; Klimek et al., 2007; Chong et al., 2009; Gullstrand et al., 2009; Massaad et al., 2009); and 3) one in the medial/lateral direction of the gait (Gordon et al., 2009).



A graphic example of the three-dimensional postitions (or rectangular components) of the BCOM during level walking at 1.39 m/s has been proposed for a male subject aged 6 to 13 (Figure 1.8), for another aged 25 to 35 (Figure 1.9) and for a last one aged 56 to 65 (Figure 1.10). All these figures are reported at the end of the chapter. Importantly, in literature, it has been widely demonstrated that, in **human walking**, the BCOM moves in a cube with 4 cm of length, width and height dimensions (as shown in Figure 1.11 sideways. Average displacement of the BCOM during level walking, in Whittle (1997)).

A graphic example of the three-dimensional positions (or rectangular components) of the BCOM during level running at 2.50 m/s has been proposed for the same male subjects aged 6 to 13 (Figure 1.12), 25 to 35 (Figure 1.13) and 56 to 65 (Figure 1.14), respectively. All these figures are reported at the end of the chapter. Importantly, in literature, it has been widely demonstrated that, in **human running**, the BCOM moves in a cube with 8 cm of length, width and height dimensions.

In effect, graphs at the end of the chapter are drawn up by means of the application Excel. The chosen independent variable was the movement time (1/100 seconds), whereas the dependent variables were the positions of the BCOM along different movement directions: a) blue for anterior/posterior direction; b) red for vertical direction; and c) dark green for medial/lateral direction. Each graph refers to only one male subject of three different age groups. Particularly, for the male subjects aged 6 to 13 and 56 to 65, 1/100 seconds from 0 to 600 were represented; for the subject aged 25 to 35 1/100 seconds from 2800 to 3400 were represented. In this way, the same size time interval has been analysed. In forward direction, graphic limits are -0.1/0.4 m (walking) and -0.3/0.4 (running) (step 0.1 m); in vertical direction, 0.7/1.1 m (walking and running) (step 0.1 m); in lateral direction, -0.05/0.30 m (walking and running) (step 0.05 m).

As shown in these figures: a) in the anterior/posterior and vertical directions, the BCOM carries out periodical movements, the period of each movement is equal to the duration of a double step (Alexander et al., 1978a; Antonsson et al., 1985; Crowe et al., 1995; Kerrigan et al., 1995; Saini et al., 1998; Tesio et al., 1998; Detrembleur et al., 2000; Gard et al., 2001; 2004; Cavagna et al., 2005; Massaad et al., 2009); b) in the medial/lateral direction, the BCOM carries out periodical movements, the period of which is equal to the duration of a single step; their frequencies are only one-half the vertical oscillations (Antonsson et al., 1985; Crowe et al., 1995; Detrembleur et al., 2000; Scheleihauf, 2004). Age-related changes in joint kinetics may also influence BCOM motion in the medio-lateral direction (Hernandez et al., 2009); c) therefore, the motion of the BCOM undergoes oscillations and it is the combined result of all complicated physiological processes and mechanical actions involved in moving (Crowe et al., 1995); d) this pattern occurs similarly in walking and running; finally e) as expected, in running, the three-dimensional displacement of the BCOM has been accelerated in comparison to walking. This pattern is particularly evident along the vertical direction (Thorstensson et al., 1984).

Moreover, these periodic three-dimensional trajectories of the BCOM, could also be graphically represented by the application *Grapher*, on a Macintosh Notebook (Grapher in Design - Wikipedia, the free encyclopedia, 2009). The three rectangular components are then plotted in a three-dimensional graph in which each component is considered aside from the movement time (Figure 1.15).

Specifically, the examples of this graphical representation are: a) on the left: three-dimensional displacement of the BCOM in downhill walking (-20%) at 0.83 m/s for a male aged 25 to 35 (front view); b) in the middle: three-dimensional displacement of the BCOM in level walking at 1.11 m/s for a male aged 36 to 45 (rear view); and c) on the right: three-dimensional displacement of the BCOM in uphill walking (+25%) at 1.94 m/s for a male aged 25 to 35 (front view).

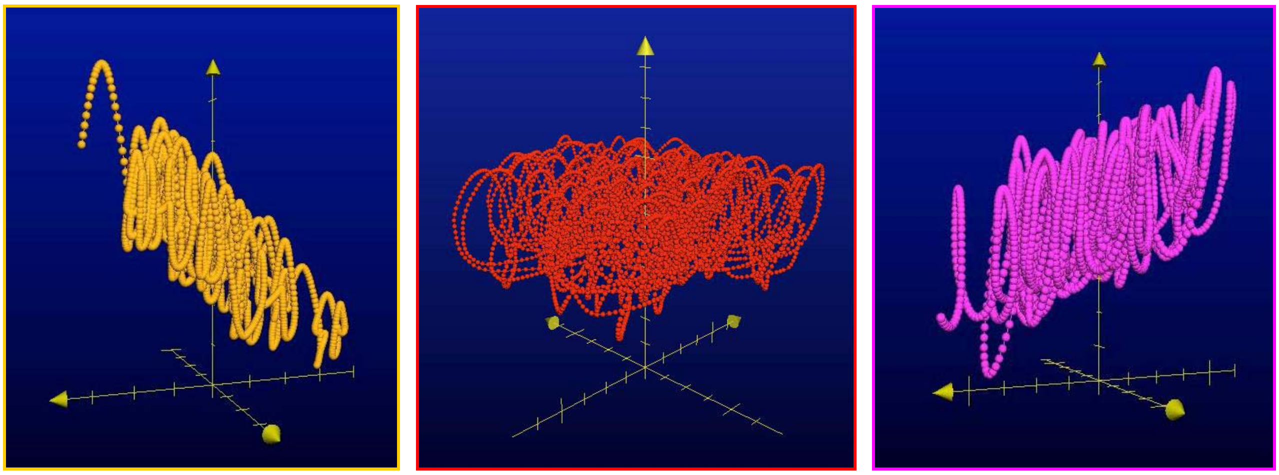


Figure 1.15. Examples of three-dimensional displacement of the BCOM as time-independent.

3.6.3. The three-dimensional movements of the body centre of mass and gait analysis

Kinematics measurements have shown that the BCOM describes a smooth sinusoidal path when projected on the sagittal plane and it also displaced laterally on the transverse plane (see also chapter 6, par. 2).

The three-dimensional double-curved trajectory which the BCOM describes during a single step could be determined as well as the velocities and accelerations of the BCOM in its trajectory. The double curved trajectory of the relative movement of the BCOM easily gives a clear picture of the absolute movement of the latter in the immobile space (McGinnis, 2005).

Precise and thorough knowledge of the movement of the BCOM is of the uttermost importance in analysing movement. Therefore, graphical examples are presented in Figure 1.16a and 1.16b.

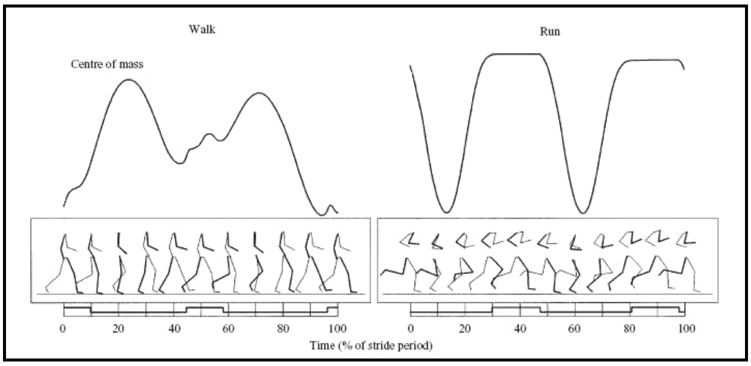


Figure 1.16a. Vertical displacement of the BCOM according to phases of the gait cycle, in Willems et al. (1995).

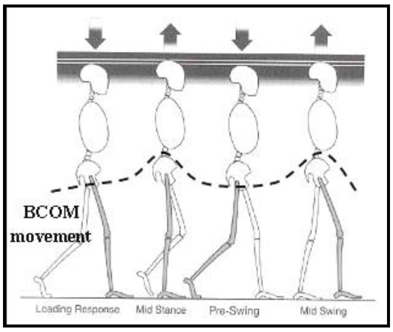


Figure 1.16b. Vertical displacement of the BCOM according to the time-variable.

It enables one to deduce the direction and magnitude of the external forces which act during walking (and running). The centre of mass of a body or a system of bodies on which any internal and external forces act, always moves as if all the masses were united in it and as if all the external forces were directly applied to it (Figure 1.17). Nevertheless, the internal forces exert no influence on the movement of the BCOM because they act in pairs, counterbalancing each other.

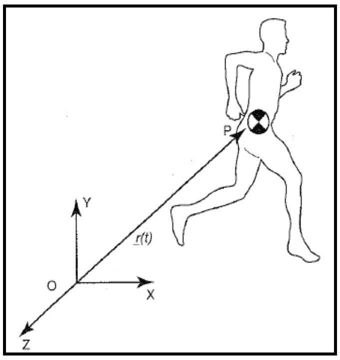


Figure 1.17. Body centre of mass idealized as a point P in a space. O is the origin of the X, Y reference frame fixed on the ground, in Scheleihauf (2004).

When taking all this into account, it is possible to answer *Yes, it is possible* to the initial question *Is it possible to use the BCOM to describe the Gait Cycle?*.

Therefore, in the following sections we will develop this issue. As a consequence, we will try to explain how the BCOM is a fundamental biomechanical body feature in gait analysis (see par. 4 below).

4. HUMAN GAIT: MECHANICAL LOCOMOTION PARADIGMS

4.1. Introduction

Legged locomotion is not particularly efficient because the limbs need to be continually repositioned on the ground, and the velocity of the foot drops to zero at each step (Alexander, 2003). The basic features of the two main modes of human progression (see par. 2.4 above) are the same: each step presents one phase of stance and one of swing. The timing of the events in the cycles are different, the stance of each foot being longer in walking and shorter in running, whilst the swing shows the opposite trend (Segers et al., 2007a; 2007b).

Despite the fact that legged locomotion is the result of the coordinated actions of dozens of muscles, each gait can be described by a simple paradigm, i.e. an analogy with a basic physical model, which will try to explain and highlight the overall mechanics of the progression (Saibene et al., 2003). Walking and running are two mechanisms for minimizing energy expenditure during terrestrial locomotion (Raynor et al., 2002; Starke et al., 2009).

Therefore, in this section, we will investigate and fully describe the simple mechanical paradigms of the two different types of locomotion, walking and running (see also Alexander, 1984; Aleshinsky, 1986a; Alexander, 2003; 2005; Geyer et al., 2006; Iida et al., 2008). Finally, we will focus on their main characteristics.

4.2. Potential energy and kinetic energy

The term *energy transformation* designates the conversion of mechanical energy from translational kinetic into rotational kinetic form and back or from kinetic into potential form and back (Raynor et al., 2002; Zatsiorsky, 2002).

First of all, it is necessary to define the mechanical energy of the BCOM (TE_{BCOM}; Scheleihauf, 2004). These concepts are in fact necessary to clearly understand the properties of the two mechanical paradigms. Mathematically, TE_{BCOM} is the sum of the **(gravitational) potential energy** (PE; Robertson et al., 2004; Winter, 2005) and the **kinetic energy** (KE; Cavagna et al., 1977; Schepens et al., 2001).

Further energy properties are also described in Winter et al. (1976); Enoka (2002); Zatsiorsky (2002); Alexander (2003); Hallemans et al. (2004); Ortega et al. (2005); Alexander (2005); Nigg et al. (2005); Cavagna et al. (2006); Segers et al. (2006) and Richards (2008). In this context, it is important to remember that: a) potential energy PE is ‘directly proportional to vertical position’; b) kinetic energy KE, on the other hand, is ‘directly proportional to the speed^2’ (Cavagna et al., 1977).

4.3. Walking

4.3.1. The pendulum-like mechanism

A first approach to understanding a complex locomotor task, such as human walking, is to understand the basis mechanics of the multi-segmented body using simple mechanical models without muscles (Zajac et al., 2003). Clearly, the less complex a walking model, the easier it is to analyze and gait insight into fundamental mechanisms.

Importantly, a human walks as if on a rimless spoked wheel, with continuous raising and lowering, accelerations and decelerations, of the body. This movement could be defined as a function of arc radius of curvature (Minetti et al., 1994; Adamczyk et al., 2006).

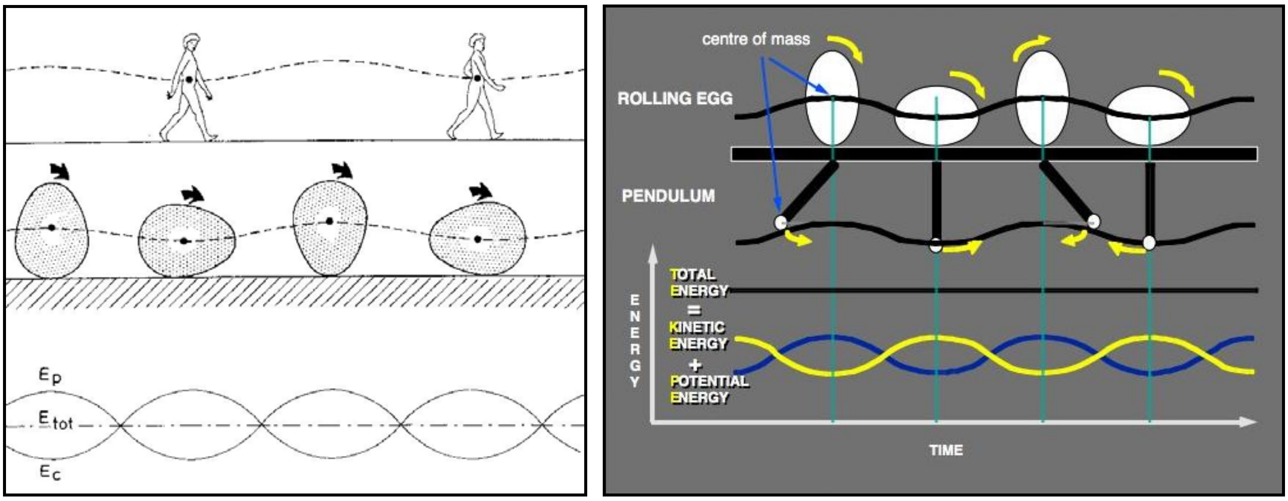


Figure 1.18. The pendulum like mechanism at the optimal walking speed, in di Prampero (1985).

Consequently, as illustrated in Figure 1.18 above, walking has been referred to as controlled falling and is similar to a **swinging pendulum** (according to Cavagna et al., 1963; Hemami et al., 1973; Cavagna et al., 1977; Alexander, 1984; di Prampero, 1985; Aleshinsky, 1986b; Cavagna, 1988; Saibene, 1990; Steudel, 1990; McKinnon et al., 1992; Diedrich et al., 1995; Heglund et al., 1995; Donelan et al., 1997; Pai et al., 1997; Cavagna et al., 2000; Fonseca et al., 2001; Cavagna et al., 2002; Donelan et al., 2002a; Raynor et al., 2002; Bastien et al., 2003; Pratt et al., 2003; Zajac et al., 2003; Gage et al., 2004; Hallemans et al., 2004; Cavagna et al., 2005; Geyer et al., 2005; Kuo et al., 2005; Ortega et al., 2005; Adamczyk et al., 2006; Geyer et al., 2006; Gottschall et al., 2006; Holt et al., 2006; Hoyt et al., 2006; Mian et al., 2006b; Segers et al., 2006; Kuo, 2007; Segers et al., 2007b; Mahaudens et al., 2008; Sawicki et al., 2008; Houdijk et al., 2009; Mahaudens et al., 2009).

This model: a) is planar; b) has completely rigid limbs, exactly one point-foot in contact with the ground at any time and instantaneous transition between legs; and c) consists of a point mass equal to the walker's body mass and a rigid strut that connects the mass to the point of ground contact (Cavagna et al., 1988).

In detail, the leg movements resemble the movements of a compound pendulum. In fact, the legs are periodically raised and lowered under the influence of gravity. Importantly, the ideal pendulum serves as an example of a conservative system: the total mechanical energy of the swinging pendulum remains constant (Zatsiorsky, 2002).

Therefore, it is important to remember that, in an ideal pendulum: a) the total mechanical energy is constant over an oscillation cycle; b) the changes in kinetic and potential energy of the body are exactly out of phase; and c) the motion is performed due to the transformation of the potential energy into kinetic energy, and *vice versa*. In detail, at each stride, the BCOM is successively behind, or in front of the point of contact with the foot on the ground. When the BCOM is behind the point of contact, the link to the ground causes a forward deceleration; as a result, it is also possible to detect a decrease in kinetic energy, a vertical rise in the BCOM, and an increase in gravitational potential energy (Cavagna et al., 2005). Usually, this phenomena happens at mid-stance (Scheleihauf, 2004). As the BCOM moves forward to the point of contact on the ground, the link to the ground provokes a decrease in the height of the BCOM and a concomitant increase in the forward speed, as some of the gravitational potential energy is converted back into the kinetic energy like an inverted pendulum (Cavagna et al., 2005; Detrembleur et al., 2005; Gottschall et al., 2006).

Let's have a look how it went about this.

4.3.2. Potential and kinetic energy in walking

In walking, potential (PE) and kinetic energy (KE) are **out of phase** (Cavagna et al., 1977; Alexander, 1984; Hreljac, 1993a; Minetti et al., 1994; Cavagna et al., 2000; 2002; Hallemans et al., 2004; Alexander, 2005; Cavagna et al., 2005; Bewiener et al., 2006; Gottschall et al., 2006; Hoyt et al., 2006; Segers et al., 2006; 2007; Sawicki et al., 2008). When potential energy is high, kinetic energy is low, and *vice versa*. Moreover, if the curves representing potential and kinetic energies are summed up numerically, the fluctuations in total mechanical energy of the BCOM are seen to be relatively small (Scheleihauf, 2004). This result indicates that the mechanical energy of the body (TE) is mostly dispersed when people are free to choose the speed at which they walk (*optimal speed*; Mansour et al., 1982; Cavagna et al., 1983; Steudel, 1990; Martin et al., 1992; Hreljac, 1993a; Minetti et al., 1993; Brisswalter et al., 1996a; Sekiya et al., 1997; Cavagna et al., 2000;

Danion et al., 2003; Bertram, 2005; Paroczai et al., 2006; Jordan et al., 2007; Mahaudens et al., 2009; Racic et al., 2009; Steudel-Numbers et al., 2009).

In effect, the centre of mass kinetic fluctuation patterns during the stance phase of both downhill and uphill walking are similar to level walking in terms of the symmetrical decreases and increases (Gottschall et al., 2006). In detail: a) during downhill walking, to maintain a constant average speed, energy must be dissipated to resist gravity. This occurs at and immediately after heel strike, and both the gravitational potential and the kinetic energy of the body centre of mass decrease (Neptune et al., 2004); and b) during uphill walking, energy must be generated to overcome gravity. This occurs prior to toe-off, and both the gravitational potential and the kinetic energy of the body centre of mass increase.

The maximum theoretical efficiency of the energetic exchange between kinetic and potential energy is only as high as 65% and varies depending on walking speed and stride frequency (Sasaki et al., 2006). Efficiency (see also chapter 20, par. 4) in walking is maintained by the effective interchange between potential and kinetic energy. In addition, the pendulum-like transfer between potential and kinetic energies reaches a maximum at the speed at which the weight-specific energy, needed to move the BCOM a given distance, is at its minimum (*optimal speed*). This speed is about 0.78 m/s at 2 years of age, and increases progressively with age up to 1.39 m/s at 12 years of age and in adults (Cavagna et al., 1983; Saibene et al., 2003; see also chapter 10).

Finally, three conditions must exist for ideal mechanical energy exchange (Gottschall et al., 2006): 1) the gravitational potential energy maximum occurs at the same time as the kinetic energy minimum and the kinetic maximum must occur at the same time as the gravitational potential minimum; 2) the magnitudes of the energy fluctuations must be the same; and 3) the gravitational potential and the kinetic fluctuations must be mirror images of each other.

As far as our study is concerning, results related to the interchange between PE and KE (and total energy) in level walking are presented in chapter 11.

4.4. Running

4.4.1. The spring-mass model

The vertical motion of the BCOM during running is compared with the oscillation of a **spring-mass system** (according to di Prampero, 1985; Williams, 1985; Cavagna, 1988; Blickhan, 1989; Steudel, 1990; Caldwell et al., 1992; Diedrich et al., 1995; Bullimore et al., 2002; Cavagna et al., 2002; 2005; Geyer et al., 2005; Morin et al., 2005; Geyer et al., 2006; Holt et al., 2006; Hoyt et al., 2006; Morin et al., 2006; Segers et al., 2006; Bullimore et al., 2007; Hunter et al., 2007; Segers et al., 2007a; 2007b; Grimmer et al., 2008; Iida et al., 2008; Bullimore et al., 2009).

As a result, this model consists of a point mass equal to the walker's body mass and a compliant spring that connects the mass to the point of ground contact (Figure 1.19). Precisely, the movement of the BCOM during running is then similar to a bouncing ball/pogo stick paradigm (Cavagna et al., 1977; Williams, 1985; Saibene et al., 2003; Bullimore et al., 2007) with two different components: 1) the vertical component; and 2) the horizontal component.

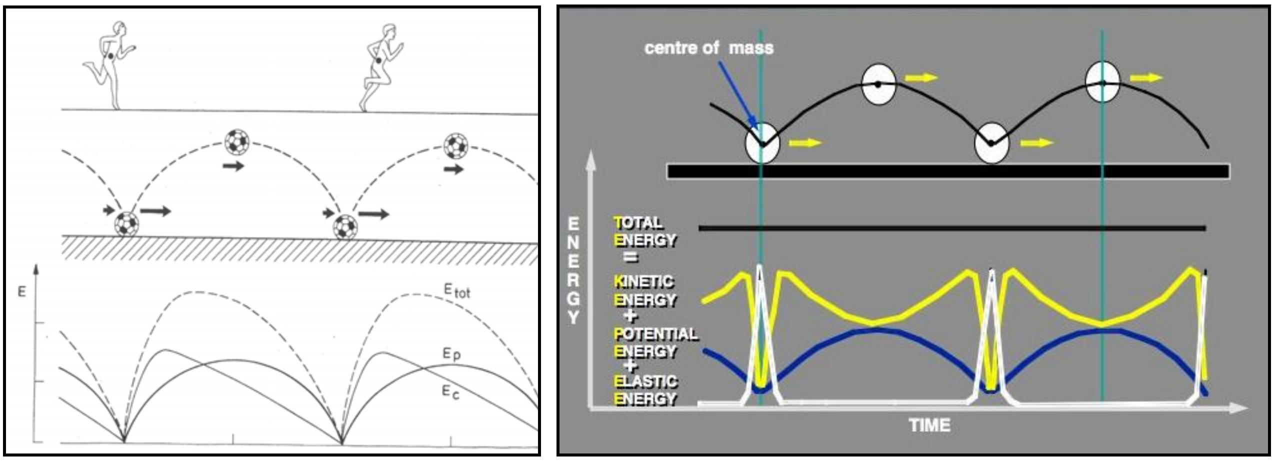


Figure 1.19. The spring-mass model at the optimal running speed, in di Prampero (1985).

Moreover, the spring-mass model assumes the same characteristics of the spring and the same height and velocity of the centre of mass at landing and take-off (Cavagna et al., 2008a). However, as the model was pointed out, it was highlight that this is a simplification because: a) the leg has to be more stiff during landing than during take-off; and b) animals in general do not have a similar take-off and landing velocity as assumed in the model. They take off with straightened legs and land with bent legs (Blickhan, 1989).

As a result, the step period and the vertical oscillation of the BCOM can be divided into two parts (according to Cavagna et al., 2008b): 1) the period during which the ground reaction force is greater than body weight (e.g. the lower part of the oscillation, taking place during the contact of the foot with the ground) (Schepens et al., 2001); and 2) the period during which the vertical ground reaction force is less than body weight (e.g. the upper part of the oscillation, taking place both during the ground contact and the aerial phase).

Alternatively, the step period and the vertical displacement were also divided into their fractions, taking place during the ground contact time and the aerial time, respectively (Cavagna et al., 2008b). During the period of absorption, the BCOM falls from its peak height during double float. The velocity of the BCOM decelerates horizontally during this period as well. After stance phase reversal, the BCOM is propelled upwards and forwards during stance phase generation. The limb is then propelled into swing phase after toe off (Williams, 1985). At swing phase reversal, the next period of absorption begins.

The spring-mass model is largely used in the interpretation of experiments aimed to determine the changes in spring stiffness and step frequency with speed, with grounds of different compliance and the effect of spring stiffness on energy expenditure. Of utmost importance, despite its relatively simplicity, the model has contributed to better understanding of running mechanics including the relationship of stiffness characteristics to stride frequency (Hunter et al., 2007).

4.4.2. Potential and kinetic energy in running

In running, potential (PE) and kinetic (KE) energy increases are **in phase** (Alexander, 1984; Williams, 1985; Cavagna, 1988; Saibene, 1990; Alexander, 2005; Cavagna et al., 2005; Bewiener et al., 2006; Hoyt et al., 2006; Segers et al., 2006; 2007a). Running has been likened to an individual pogo stick, propelling oneself from a low point during the stance phase reversal to a peak during double float. Kinetic and potential energy peak in mid-swing. As the foot comes into contact with the ground, kinetic energy is lost. As the BCOM falls towards the ground, potential energy is lost.

Much of the lost kinetic and potential energy is converted into elastic potential energy (EL) and stored in the muscles, tendons and ligaments. In order to clarify this concept, we will briefly present the main properties of this parameter.

As far as our study is concerning, results related to the interchange between PE and KE (and total energy) in running are presented in chapter 11.

4.4.3. Elastic energy in running

Elastic energy utilization that stores and returns mechanical energy is considered to be an important metabolic energy saving mechanism, especially in running (Asmussen et al., 1974; Alexander, 1984; 1988; Caldwell et al., 1992; Brisswalter et al., 1996; Ettema, 1996; Alexander, 2005; Cavagna et al., 2005; Bewiener et al., 2006; Sasaki et al., 2006; Bullimore et al., 2007; Whittington et al., 2008). These elastic storage and return of energy is directly related to stiffness of the running leg (Hunter et al., 2007).

To be specific, the running mechanism used by human adults is characterized by an elastic rebound of the body at each step (EL = elastic energy; Cavagna et al., 1977; Cavagna, 1988; Cavagna et al., 2005; 2008a). Muscles, tendons and ligaments can all behave as springs, storing elastic energy (especially in the Achilles tendon; Sasaki et al., 2006) when they are stretched and returning it when they recoil.

A stretch-rebound cycle of the muscle-tendon units takes place at each running step with a large input and output of mechanical energy during the vertical oscillation of the bouncing system. The storage and utilization of elastic energy in running is not as efficient as the conversion of gravitational potential energy into forward kinetic energy in walking (Scheleihauf, 2004).

In particular, as in a pogo-stick, part of the total mechanical energy of the system during the flight phase is transformed into elastic energy during the first half of the contact phase, via tendon stretch. In the second half of the phase, a consistent part of the stored energy is propelled back into the system via tendon recoil, in preparation for the next stride (Saibene et al., 2003).

The elastic-system utilization is greater in running above than below the preferred walk-run transition speed (Thorstesson et al., 1987; Hreljac, 1993a; 1993b; 1995a; 1995b; Brisswalter et al., 1996; Raynor et al., 2002; Rotstein et al., 2005; Sasaki et al., 2006; De Smet et al., 2009; Jordan et al., 2009).

4.4.4. The pendulum-like mechanism *versus* the spring-mass model

Each gait can be described by a single paradigm which helps understand the overall mechanisms of the progression along the ground (Saibene et al., 2003). In fact, inverted pendulum models of walking and mass-spring models of running are powerful in understanding the transition from one gait to another (Zajac et al., 2003).

However, inverted pendulum and mass-spring models do not attempt to account for the multi-linked nature of the legs and provide little insight into muscle coordination principles.

Importantly, although some amount of elastic strain energy is stored and reutilized in each cycle of running, the single-pendulum model is less efficient than in walking (Cavagna et al., 1977; Saibene et al., 2003). Furthermore, the bouncing mechanism of running, contrary to the pendulum-like mechanism of walking, requires much greater muscular intervention to maintain the motion of the BCOM (Cavagna et al., 2008b).

Running efficiency is primarily maintained in two ways: 1) the storage and return of elastic potential energy by the stretching of elastic structures (especially tendons). Therefore, the storage of energy in elastic structures of the lower extremities plays a more important role in running and sprinting than in walking (Novacheck, 1998); and 2) the transfer of energy from one body segment to another by two joint muscles such as the rectus femoris and the hamstrings. These two mechanisms do not occur without some cost to the system.

The efficiency in running has been calculated as about 40-50% (Cavagna et al., 1964). Such a high value (Cavagna et al., 1977) involves a contribution of a substantial amount of energy delivered at a very low cost. Finally, this appears to be identified as elastic recoil energy from the stretched contracted muscle and amounts to about half the energy spent in running. Due to this, the conventional wisdom is that the walk-run transition is not determined by the mechanical limitations of the system, but rather by an inherent need to conserve energy (Raynor et al., 2002).

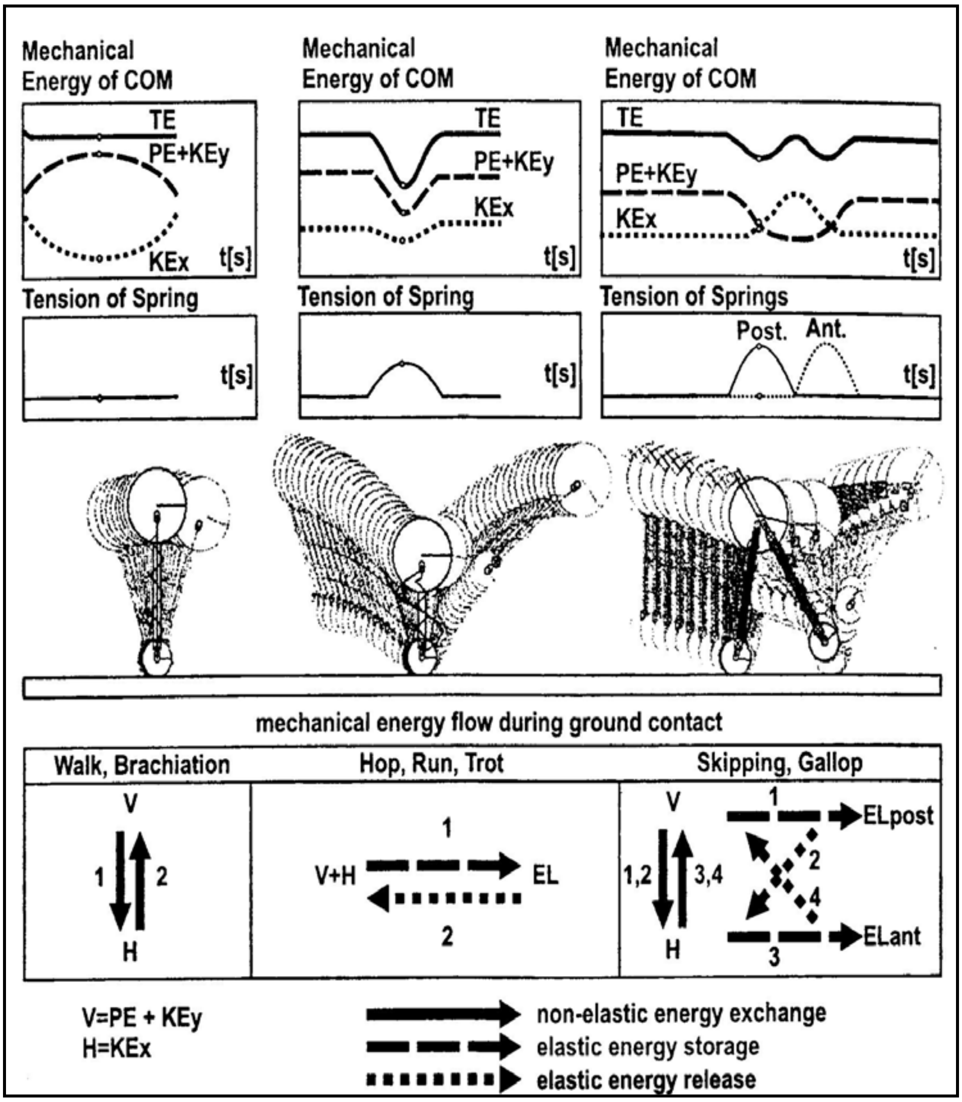


Figure 1.20. The main mechanical paradigms of terrestrial locomotion, in Saibene et al. (2003).

Skipping, an interesting gait by children aged 4 to 5 (Minetti, 1998; Saibene et al., 2003), by astronauts moving on the Moon, by quadrupeds during galloping and by crows is a combination of walking and running in a single stride (see Figure 1.20 above). As a result, the mechanical paradigm is a system made of two pogo-stickes linked together by sharing the upper edge.

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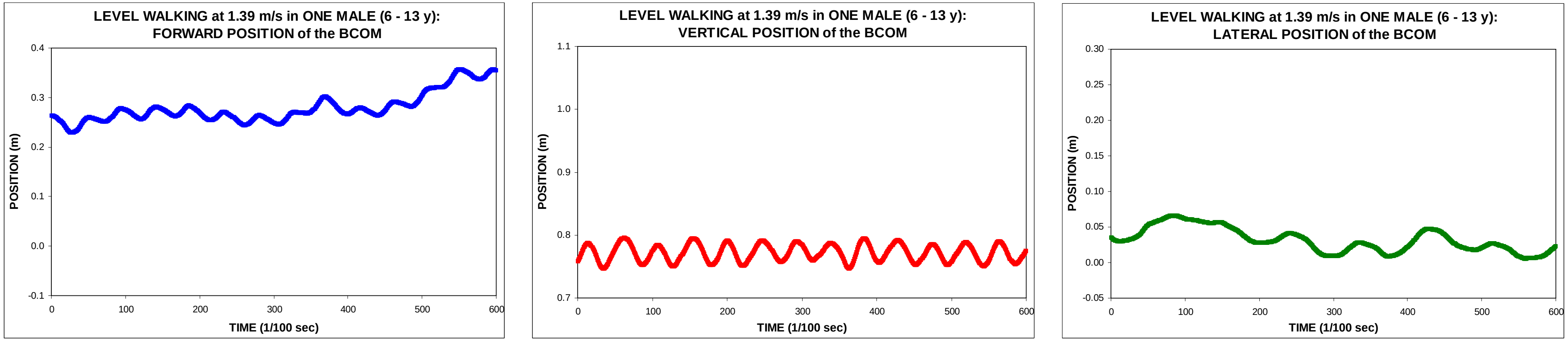


Figure 1.8. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 6 to 13, in level walking at 1.39 m/s.

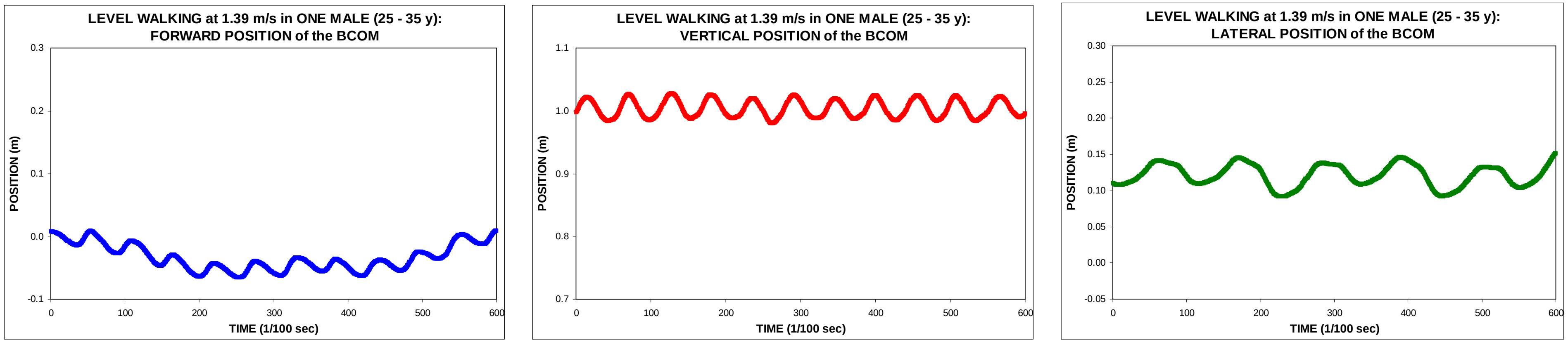


Figure 1.9. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 25 to 35, in level walking at 1.39 m/s.

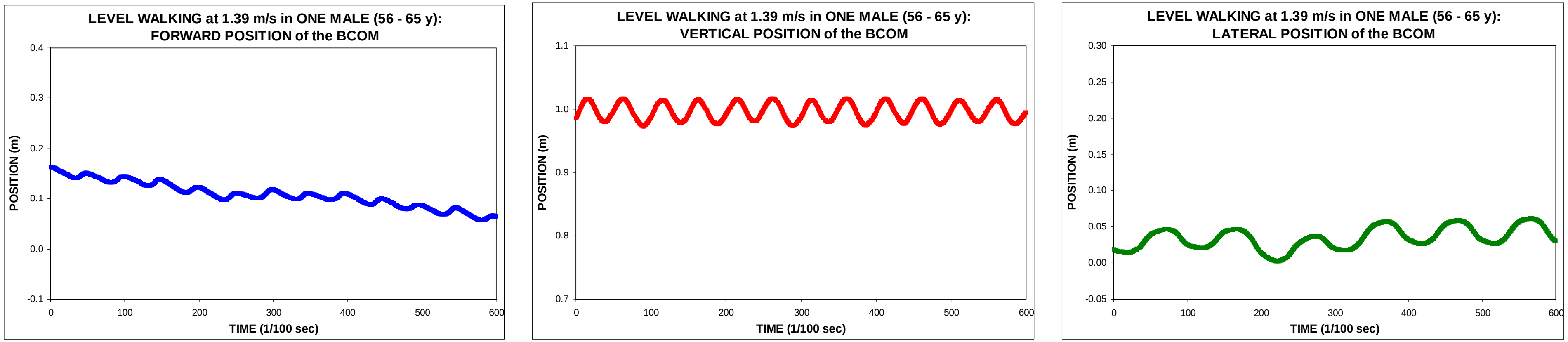


Figure 1.10. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 56 to 65, in level walking at 1.39 m/s.

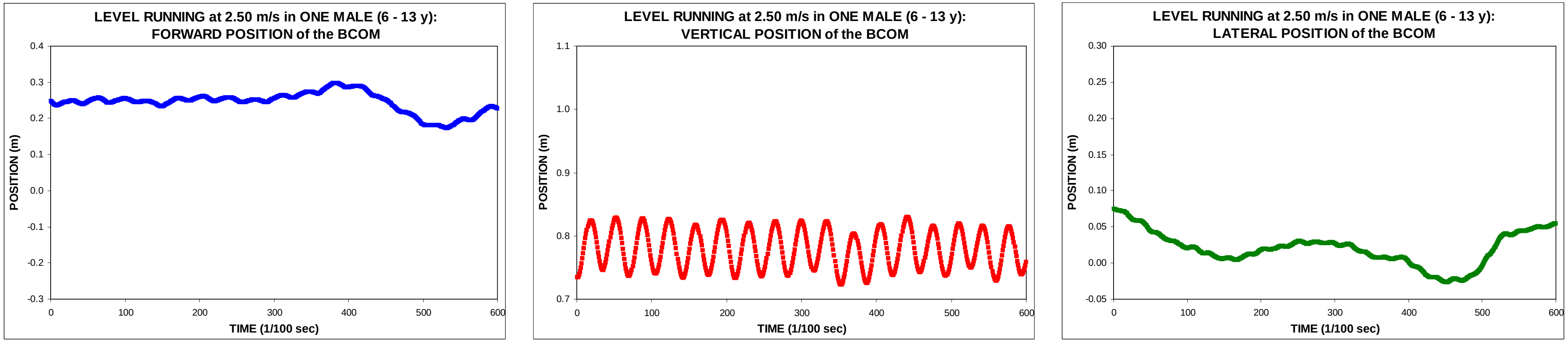


Figure 1.12. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 6 to 13, in level running at 2.50 m/s.

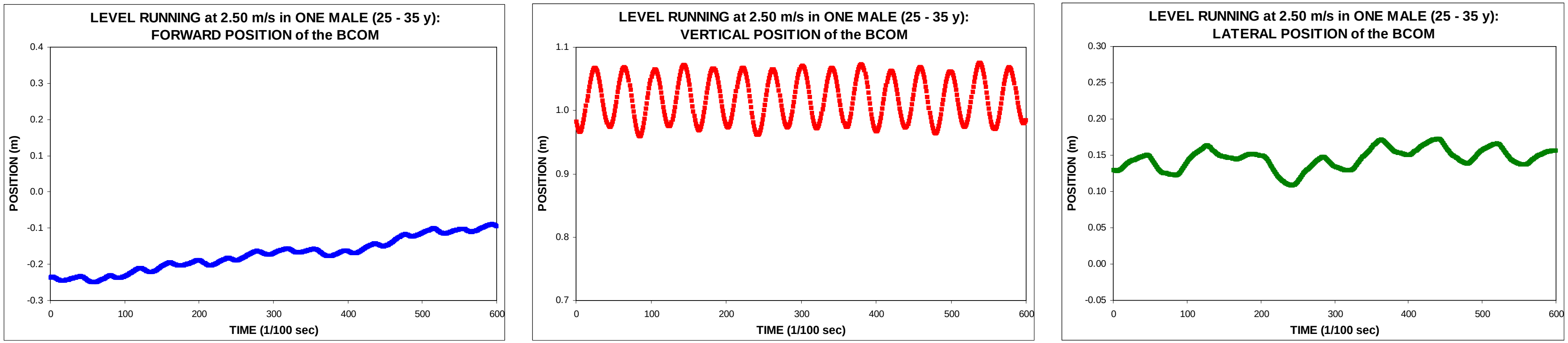


Figure 1.13. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 25 to 35, in level running at 2.50 m/s.

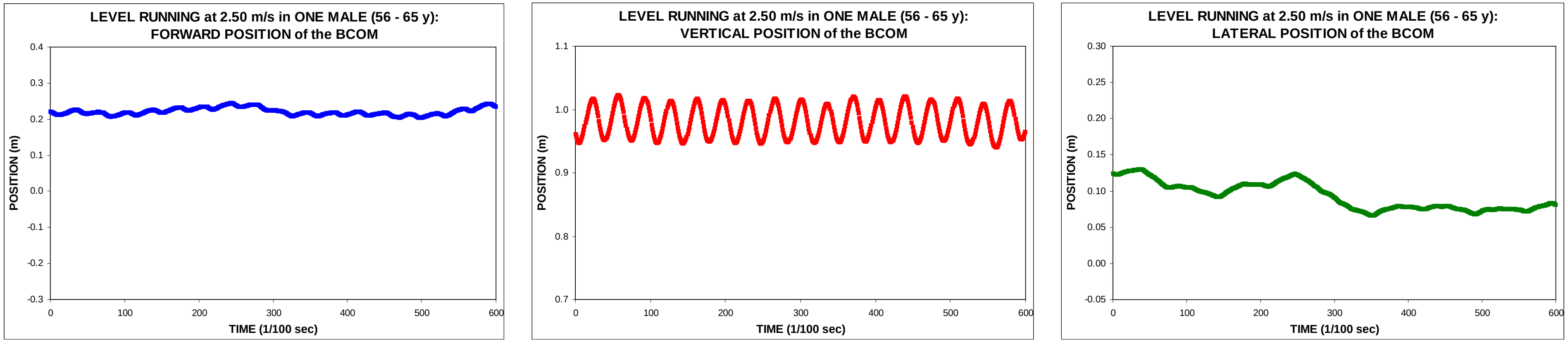


Figure 1.14. Anterior/posterior (on the left), vertical (in the middle) and medial/lateral (on the right) position of the BCOM, for a male subject aged 56 to 65, in level running at 2.50 m/s.

Chapter 2

BODILY SYMMETRIES AND RUNNING ECONOMY

1. SYMMETRY: SOME DEFINITIONS

In general, symmetry is 'the balanced arrangement of body parts or shapes around a central point or axis' (Symmetry in Biology - Wikipedia, the free encyclopedia, 2009). As a result, that is, the size, shape, and relative location on one side of a dividing line mirrors the size, shape, and relative location on the other side.

Some examples of this general symmetry are illustrated in Figure 2.1.

Figure 2.1. Some examples of general symmetry.

Symmetry essentially reflects *order*, *harmony*, *equilibrium* (Møller, 1997; Penton-Voak et al., 2001; Koehler et al., 2004; Rhodes et al., 2007; Little et al., 2008).

However, in biology, symmetry is approximate (i.e. plant leaves, while considered symmetric, will rarely match up exactly when folded in half). Furthermore, symmetry may refer only to the external form and not to the internal anatomy.

To better understand the significance of the symmetry and its behaviour with respect to running economy, it makes important to define some terms focusing on their proper meanings.

To avoid confusion, different concepts have to keep in mind.

1. **Bilateral symmetry** (Figure 2.2). Many humans and animals have a body form that is bilaterally (*bi* = two, *latus* = side) symmetrical (Allard et al., 2009). In detail, this means that their body could be divided into matching halves by drawing a central axis or a line down the centre (e.g. human faces, arthropods that are built like humans, fossil evidence, leaves of most plants, insects,

spiders, worms and many other invertebrates and so on). Indeed, this constitutes an indicator of developmental stability (Penton-Voak et al., 2001; Yost, 2009).

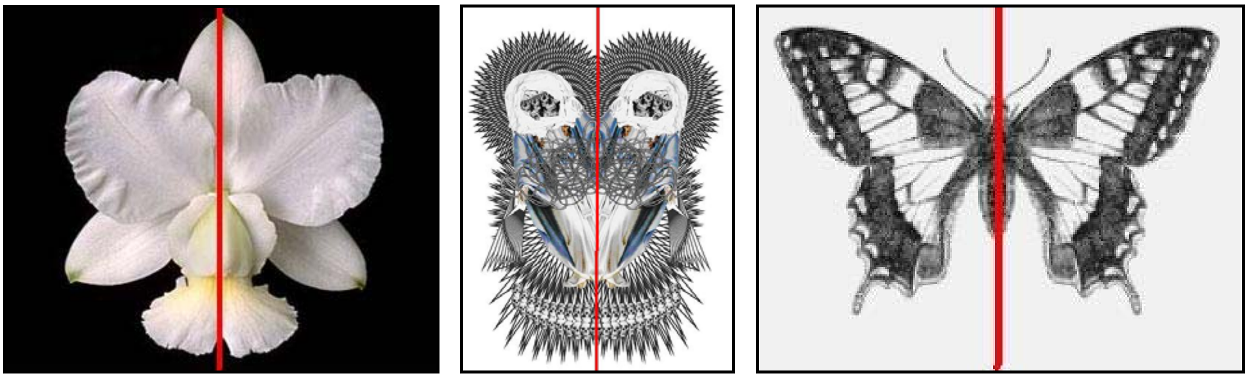


Figure 2.2. Some examples of bilateral symmetry.

In bilateral symmetry (also called *plane symmetry*), only one plane (the sagittal) will divide an organism into roughly mirror image halves (with respect to external appearance only).

Thus, there is approximate reflection symmetry. Often the two halves can meaningfully be referred to as the right and left halves, e.g. in the case of an animal with a main direction of motion in the plane of symmetry. An example would be an airplane, whereby a plane passing through the center of the plane from tip to tail would divide the plane into two equal parts (on external surface).

However, it has been demonstrated how the vertical symmetry is most salient, followed by horizontal and then oblique orientations (Herbert et al., 2002). Furthermore, it has been shown how bilateral variations in dimensions of upper and lower limb bones are attributable to difference in mechanical stress and strain that the bones are subjected to during bone growth (Kanchan et al., 2008). Probably, this skeletal asymmetry in the upper limbs is usually prominent on the dominant side while in the lower limbs on the other side, possibly due to supportive contra lateral muscle contractions, that influence the bone growth.

Figure 2.3a. An example of asymmetrical and symmetrical faces in humans (courtesy of Elena Seminati).

Otherwise, especially in human faces (Figure 2.3a above), there are some important situations in which the aforementioned bilateral symmetry is not so easy to identify (Grammer et al., 1994; Penton-Voak et al., 2001; Koehler et al., 2004; Rubenstein, 2005; Jones et al., 2007; Lopez-Garcia et al., 2007; Rhodes et al., 2007; Little et al., 2008; Komori et al., 2009).

As shown in Figure 2.3b below, of course, the bilateral symmetry of vehicles is not perfect: the car in the picture, for example, has a mirror only on the driver's side, which destroys what would otherwise be perfect bilateral symmetry. Herein symmetry was achieved by reflecting the driver's side of the car. If we reflect the other side, there will be no mirror and, although this car is just as symmetrical as the one with two mirrors, it will not comply with motor vehicle codes.

Figure 2.3b. Imperfect bilateral symmetry in cars.

2. **Radial symmetry** (Figure 2.4). In radial symmetry, all planes passing through a central axis (normally vertical) divides the form into two identical halves that are mirror images of each other (Symmetry in Biology - Wikipedia, the free encyclopedia, 2009).

Figure 2.4. Some examples of radial symmetry.

Such a form will have distinct ends (usually top and bottom) and any plane that passes through its longitudinal axis (a line from end to end through the center) will create two similar halves. An organism with radial symmetry exhibits no left or right sides. They have a top and a bottom surface only. Some examples are sea anemone, floating animals such as jellyfish, slow moving organisms such as sea stars and many flowers, such as buttercups and daffodils.

3. **Spherical symmetry** (Figure 2.5). In spherical symmetry, any plane that passes through the center of the object divides the form into two identical halves that are mirror images of each other (Symmetry in Biology - Wikipedia, the free encyclopedia, 2009).

Figure 2.5. Some examples of spherical symmetry.

Some examples are objects are shaped like spheres or globes: specifically, they live in the animal kingdom.

4. **Laterality determination** (Roether et al., 2008). This embryonic process, which is beginning to yield its universal molecular basis, is the major effect of evolutionary developmental modification (i.e. the abolition of the visceral symmetry). Therefore, it could be concluded that laterality determination is not probably responsible for another type of biological phenomenon designed fluctuating asymmetry (Opitz et al., 2001).

5. **Fluctuating asymmetry** (Figure 2.6). This represents the deviation from perfect bilateral symmetry caused by environmental stresses, developmental instability and genetic problems during development (Valen, 1962; Leary et al., 1989; Leung et al., 1996; Opitz et al., 2001; Polak, 2003; Al-Eisa et al., 2004; Benderlioglu et al., 2004; Koehler et al., 2004; Little et al., 2008; Miller et al., 2008).

In other words, fluctuating asymmetry results from an organism's failure to cope with various inclement environmental and genetic factors (Penton-Voak et al., 2001; Little et al., 2008) or measures deviations from the ideal state of symmetry, and is therefore thought to reflect the level of genetic and environmental stress experienced by individuals or populations during development (Tomkins et al., 2001).

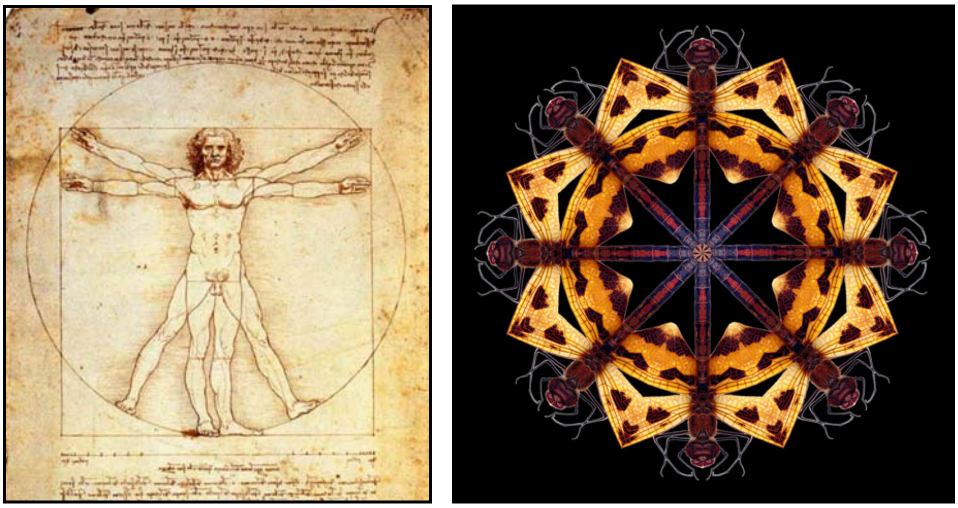


Figure 2.6. Some examples of fluctuating asymmetry.

It is thought that the more perfectly symmetrical an organism is, the better it has been able to handle developmental stress and has more developmental stability. Finally, fluctuating asymmetry could be used as an indicator of individual quality in studies of natural and sexual selection and as a bio-indicator tool for environmental monitoring and conservation biology (Penton-Voak et al., 2001; Tomkins et al., 2001).

2. ANIMAL MODELS

The literature research has showed a quite significant relationship between static anatomical and kinematic functional symmetries and running economy only in animals.

Particularly, many documented studies regard the possible relationship between 1) (a)symmetry and sexual characters or reproduction, 2) (a)symmetry and health, and 3) (a)symmetry and physical performance. In order to focus on single main topics and aspects, some of them will be briefly summarized and discussed.

2.1. Symmetry and sexual characters/reproduction

In the past, mature peacocks and their eyes (or ocelli) were studied in order to define train symmetry based on the assumption that train symmetry was the number of ocelli on the side with the greatest number minus the number of ocelli on the side with the least (Manning et al., 1991). It has been suggested that female animals tend to prefer mates who have symmetrical bodily characteristics. Moreover, by statistical analysis, it has been found that the train symmetry was very strongly correlated with the number of ocelli per train.

Subsequently, the degree of fluctuating asymmetry and the incidence of aberrant secondary sexual characters in the monogamous barn shallow *Hirundo Rustica* in two areas have been investigated, as well (Møller, 1993). It has been found that: a) the level of fluctuating asymmetry is different only in some characters (in males); and b) the incidence of aberrant feather morphology slightly depends on the geographic area.

More recently, both directional and fluctuating asymmetry in two wing traits within both sexes of the damselfly *Calopteryx Maculata* have been examined (Pither et al., 2000). It has been noted that patterns of asymmetry are generally consistent among sexes and sites (right and left), although males tend to exhibit more pronounced directional asymmetry (i.e. compensatory development).

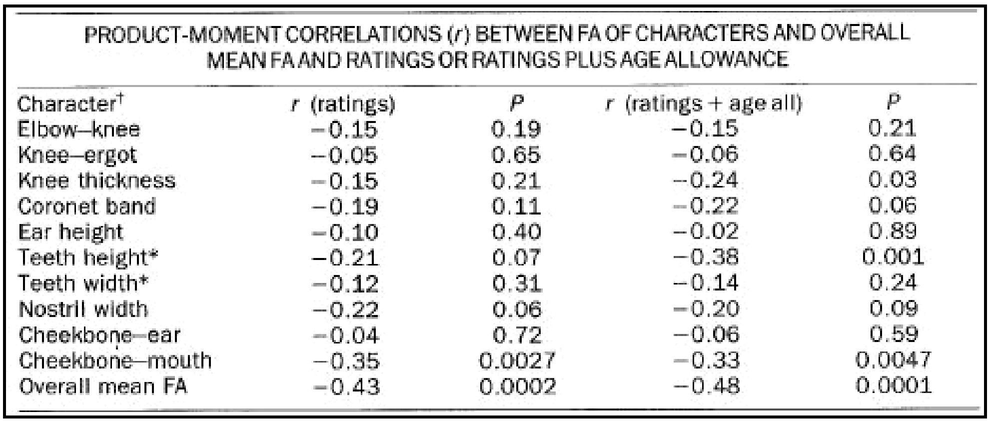
2.2. Symmetry and health

The expression of fluctuating asymmetry in some teeth of the upper jaw of gorillas has been examined (Manning et al., 1994a). Particularly, it has been demonstrated a sustained environmental deterioration primarily because of the correlation with a) year of collection and b) acquisition of the specimen. These findings supported the thesis that sexually selected structures are sensitive to environmental stress.

Fluctuating asymmetry is a promising measure of *animal welfare* (Knierim et al., 2007), as well. Therefore, its considerable potential as a welfare indicator makes it worthwhile to pursue more intensely validation studies as well as applied studies.

2.3. Symmetry and physical performance

Importantly, it has been verified that the fluctuating asymmetry in thoroughbred racehorses had an effect on racing ability and may therefore be a predictor of future performance in young horses (Manning et al., 1994b). Precisely, ten paired characters (4 in the thighs and 6 in the head) on 73 flat-racing thoroughbreds have been measured (Table 2.1).



PRODUCT-MOMENT CORRELATIONS (*r*) BETWEEN FA OF CHARACTERS AND OVERALL MEAN FA AND RATINGS OR RATINGS PLUS AGE ALLOWANCE

Character[†]	*r* (ratings)	*P*	*r* (ratings + age all)	*P*
Elbow–knee	−0.15	0.19	−0.15	0.21
Knee–ergot	−0.05	0.65	−0.06	0.64
Knee thickness	−0.15	0.21	−0.24	0.03
Coronet band	−0.19	0.11	−0.22	0.06
Ear height	−0.10	0.40	−0.02	0.89
Teeth height*	−0.21	0.07	−0.38	0.001
Teeth width*	−0.12	0.31	−0.14	0.24
Nostril width	−0.22	0.06	−0.20	0.09
Cheekbone–ear	−0.04	0.72	−0.06	0.59
Cheekbone–mouth	−0.35	0.0027	−0.33	0.0047
Overall mean FA	−0.43	0.0002	−0.48	0.0001

Table 2.1. 10 paired characters measured on thoroughbreds, in Manning et al. (1994).

Particularly, relative asymmetry was calculated ‘as right minus left, divided by the mean left and right character value’ (see also chapter 17, par. 1). To eliminate inter-observer variation, all measurements were made by only one operator (see also chapter 17, par. 1.2.2). Characters were measured twice in ten individuals. In such a way, 'repeatability ranged from 0.97 to 1.00'.

The abilities of racehorses have been estimated by handicappers. The results showed that relationships between ratings and fluctuating asymmetry are all negative (particularly for features of the head) (Figure 2.7).

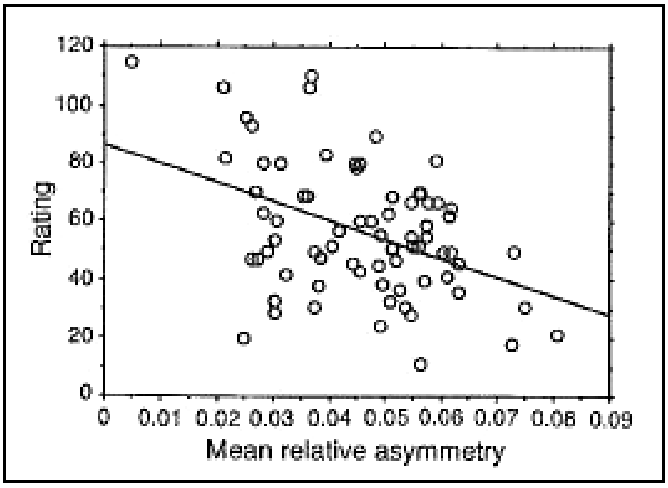


Figure 2.7. Relationship between the racing ability of 73 thoroughbread racehorses, in Manning et al. (1994).

Therefore, the estimation of overall fluctuating asymmetry may provide an additional tool which could be used, with due correction for age, in the process of selecting the best performer.

In another research, it was supposed that fluctuating asymmetry could be advocated as the preferred measure of developmental instability and as a reliable indicator of the quality (Tuyttens et al., 2005). Therefore, an experiment on weaned rabbits which were housed in different pens (welfare-friendly or barren) has been conducted. Rabbits housed in the welfare-friendly pens were less asymmetrical and the fluctuating asymmetry was negatively correlated with best performance data gain only in rabbits from the barren pens. This research demonstrated that the application of fluctuating asymmetry is an indicator of animal welfare and performance.

3. HUMAN MODELS

A limited number of studies have tried to show the same possible relationship in human models.

Particularly, only few documented studies regard the possible relationship between 1) (a)symmetry and sexual characters or reproduction and 2) (a)symmetry and healthy. Therefore, some of them will be briefly summarized and discussed in the following sections.

3.1. Symmetry and sexual characters/reproduction

In a first study, it has been demonstrated that both averageness and symmetry in faces would be preferred (Grammer et al., 1994). In a second research, it has been verified that the magnitude of the negative correlation between fluctuating asymmetry and success related to sexual selection is greater for males than for females (Møller et al., 1998).

Quite recently, it has been stated how mammographic interpretation often uses symmetry left and right breasts to indicate the site of potential tumor masses (Alterson et al., 2003). The associations between masculine facial features and non-facial body symmetry as well as facial symmetry (males and females) have been examined, too (Gangestad et al., 2003). The findings demonstrated that a component of facial features which discriminates the sexes and reflects masculinization of the face significantly co-varied with symmetry in men. Furthermore, an explicit measure of facial fluctuating asymmetry and measures of facial masculinity/femininity have been used to test the relationship between masculinity and symmetry, two putative signs of mate quality, in male faces (Koehler et al., 2004). It was found no significant correlations between facial masculinity and any other measures of asymmetry or ratings of symmetry in males.

Nowadays, it has been sought to establish whether the attractive bodies signal low fluctuating asymmetry to observes, and, if so, which aspects of attractive bodies are more predictive of lower fluctuating asymmetry (Brown et al., 2008). Strong negative correlations between fluctuating asymmetry and bodily attractiveness (both sexes) have been found. Further, sex-typical body size and shape characteristics were treated as attractive and correlated negatively with fluctuating asymmetry. In addition, by studying effects of averageness and symmetry on the judgment of facial attractiveness, it has been verified that a) for males, symmetry and averageness, while b) for females only averageness affected attractiveness ratings positively (Komori et al., 2009).

Human perceptions of facial asymmetry are driven by fluctuating asymmetries (Rhodes et al., 2009). In general, in humans fluctuating asymmetry has been shown to be related to: a) body weight (Manning et al., 1994a); b) age (Wilson et al., 1996); c) running speed (Manning et al., 1998a; 1998b); and d) developmental instability (Dongen, 2006).

3.2. Symmetry and health

In a research on children (boys and girls), it has been demonstrated that: a) small deviations from perfect bilateral symmetry is negatively correlated with health and positively correlated with sexual selection in human adults; b) accumulation, persistence and fitness implications of asymmetries during childhood are largely unknown; c) fluctuating asymmetry of the legs tended to be related; d) legs may show high developmental stability resulting from selection for mechanical efficiency; e) boys have significantly lower fluctuating asymmetry than girls and this effect resides mainly in the elbows; and f) there were significant positive relationships just between composite fluctuating asymmetry and age, height and weight (Manning et al., 1998b).

Prenatal stressors contribute to the imprecise expression of symmetrical phenotypes and display of agonistic behaviour in children and adults (Benderlioglu et al., 2004). The hypothesis that high fluctuating asymmetry would be associated with elevated levels of human reactive aggression has been verified, as well.

In a more recent study, the relationship between developmental instability and exposure to prenatal stresses using fluctuating asymmetry has been tested, too (Hall et al., 2008). Particularly, differences among individuals reflect variation in their exposure to and ability to accommodate for stresses experienced during development.

Finally, it has been verified how the general factor of mental ability (i.e. intelligence) may reflect general biological fitness (i.e. fluctuating asymmetry) (Bates et al., 2006). Indeed, the intelligence is a reflection of a more general fitness factor influencing growth and maintenance of all bodily systems, with brain function being an especially sensitive indicator of this fitness factor.

3.3. Symmetry and physical performance

In current literature, there is no evident researches focusing on this important topic. Therefore, one the main purpose of our research is the validation of such a relationship in humans (i.e. runners) featuring different running abilities, as successfully demonstrated in horses (Manning et al., 1994b).

4. RUNNING ECONOMY

The study of running economy may provide insight into mechanisms underlying economical human locomotion (Allen et al., 1985; Daniels, 1985; Morgan et al., 1992; Brisswalter et al., 1996; Craib et al., 1996; Seyfarth et al., 2002; Albracht et al., 2006; Yamamoto et al., 2008).

In effect, running economy is defined as the metabolic power per unit body mass required to run at a certain speed (Conley et al., 1980; Plews, 2000) or, equivalently, as the amount of metabolic energy needed to displace a unit of body mass over a certain distance (Martin et al., 1992; 1993; Saunders et al., 2004a; Vuorimaa, 2005; Abe et al., 2007; Scholz et al., 2008; Fletcher et al., 2009; Gullstrand et al., 2009; Running Economy in Physiology - Wikipedia, the free encyclopedia, 2009).

Thus, at any given speed, better running economy refers to a smaller rate of energy consumption (Daniels, 1985; Pate et al., 1992; Craib et al., 1996; Jensen et al., 1999; Millet et al., 2000; Hausswirth et al., 2001; McCann et al., 2003; Minetti, 2004; Abe et al., 2007; Foster et al., 2007; Capelli et al., 2008; Scholz et al., 2008; Chen et al., 2009).

Running economy is affected by both physiological factors (Kaneko, 1990; Gullstrand et al., 2009): a) age (Daniels, 1985; Pate et al., 1992); b) fluctuations in core temperature (Daniels, 1985); c) heart rate; d) ventilation (Pate et al., 1992); e) lactate; f) muscle fibre type; g) wind resistance; h) fatigue (Daniels, 1985; Morgan et al., 1992; Pate et al., 1992; Saunders et al., 2004b) and i) environment (i.e. temperature, altitude, running surface; Daniels, 1985), and biomechanical factors (Kaneko, 1990; Plews, 2000; Gullstrand et al., 2009): a) height; b) weight (Daniels, 1985; Pate et al., 1992); c) ponderal index; d) body fat; e) leg morphology; f) elastic energy (Morgan et al., 1992; Pate et al., 1992; Saunders et al., 2004b; Heise et al., 2008; Trehearn et al., 2009); g) a freely chosen stride frequency and length (Dallam et al., 2005); h) less vertical oscillation (Williams et al., 1987); i) poor running technique (Kyrolainen et al., 2001); j) peak in ground reaction time (Heise et al., 2001); k) muscle stiffness and the resonant frequency of the propulsive leg (Plews, 2000) and so on (Dallam et al., 2005; Schucker et al., 2009).

Finally, some studies on running economy have revealed that running economy is an important determinant of running performance (Conley et al., 1980; Daniels, 1985; di Prampero, 1985, 1986; Williams et al., 1987; Unnithan et al., 1990; Pate et al., 1992; Anderson, 1996; Kyrolainen et al., 2000; Hausswirth et al., 2001; Saunders et al, 2004b; Dallam et al., 2005; Vuorimaa, 2005; Abe et al., 2007; Bushnell et al., 2007; Hunter et al., 2007; McCann et al., 2008; Schucker et al., 2009).

Consequently, some interventions to improve running economy have been proposed: a) altitude exposure and training in the heat (Franch et al., 1998; Plews, 2000; Saunders et al., 2004b; Vuorimaa, 2005; Bonacci et al., 2009); b) strength training (Plews, 2000; Beneke et al., 2005; Vuorimaa, 2005; Yamamoto et al., 2008; Bonacci et al., 2009); c) the improvement of effective force (Candotti et al., 2007); d) high intensity running (Foster et al., 2007); e) a short contact time (Vuorimaa, 2005; Nummela et al., 2007; Bonacci et al., 2009); f) plyometrics, high intensity interval and altitude/hypoxia training (McCann et al., 2008; Bonacci et al., 2009); g) stretching and relaxation exercises (Vuorimaa, 2005; Heise et al., 2008; Bonacci et al., 2009; Trehearn et al., 2009); and h) rest and recovery methods (i.e. massage, physiotherapy and restoration treatment; Vuorimaa, 2005).

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UNIVERSITY OF VERONA

Department of Neurological and Visual Sciences

PhD Program in Exercise and Human Movement Science

Cycle XXII°

FIRST STUDY

HUMAN LOCOMOTION: SYMMETRY IN THE 3D TRAJECTORY OF THE BODY CENTRE OF MASS

Chapter 3

OVERVIEW OF FIRST STUDY

New biomechanical approaches will provide a different viewpoint to the mechanical paradigm of locomotion (Alexander, 2003). A series of pilot experiments and analyses showed that a series of strides in the same individual under the same condition is quite consistent, suggesting the presence of a locomotory signature. Preliminary experiments showed that, by capturing via motion capture the movement of body segments and by calculating the trajectory of the BCOM from them, the proposed methodology gives a very accurate description of the kinematics of the average stride for each individual and condition (Minetti, 2006).

Consequently, our hypothesis is that at every gait (walking and running, in particular), speed and gradient, a unique 3D *contour* can be associated (Bianchi et al., 1998; Lee et al., 1998; Detrembleur et al., 2000; Hausdorff, 2005; Minetti, 2006), where 3D *contour* is a global index of the BCOM dynamics. Differences will be detected between gender and across ages, as well (Barrett et al., 2008; Hernandez et al., 2009; Minetti, 2009).

In our research project, the establishment of a mathematical method (or function; Feng et al., 2009) and a valid evaluation protocol for the study of the BCOM pattern (three-dimensional displacement) during human locomotion is then related to: a) **gender**: male and female (Hageman et al., 1986; Barrett et al., 2008; Chumanov et al., 2008; Røislien et al., 2009); b) **age**: from 6 to 65 years (Gabell et al., 1984; Bendall et al., 1989; Kang et al., 2008; Hernandez et al., 2009; Røislien et al., 2009); c) **different gaits** (or types of locomotion): walking and running; d) **speed** (Murray et al., 1966; Bendall et al., 1989; Røislien et al., 2009); and e) **gradient**: level, uphill and downhill condition (Minetti et al., 1993).

Consequently, we studied seventy people from different age groups, ranging from 6 to 65 (mean step 10 years) who volunteered for the study, were informed and gave their full consent prior to taking part to the tests. The subjects were categorised into 7 groups, with 5 males and 5 females *per* group. These participants had to have no impediments as far as neurological or musculoskeletal pathologies affecting gait were concerned (for more details, see chapter 5).

Kinematic data was obtained by recording treadmill locomotion (walking and running) with the motion capture technique (Vicon MX system; for more details, see chapter 4). Each subject carried out these two types of locomotion on the level gradient, at 10 different speeds (from 0.83 to 1.94 m/s for walking, step 0.28 m/s; and from 1.94 to 3.06 m/s for running, step 0.28 m/s) so that each subject performed 10 trials (700 conditions, in total). In addition to this minimum general protocol, males and females aged 25 to 35 carried out walking and running at the same speed, but on different gradients (uphill: from +5 to +25%, step 5%; and downhill: from -5 to -25%, step 5%). To sum up, each subject from this group performed 100 trials (1100 conditions, in total; for more details, see chapter 5).

All testing was carried out utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

Each kinematic data has been elaborated by means of a custom-written LabVIEW (NI, USA) software (Minetti et al., 1993). In this way, we obtained information about the mathematical three-dimensional displacement (anterior/posterior, vertical and medial/lateral) of the BCOM. Most of human motion is periodic, as reflected in changes in joint angle and vertical displacement trajectories. Thus, some functions involving motion are described using transformations representing the spatial-temporal characteristics of these trajectories. Examples of these human motions are walking and running which repeat themselves periodically.

The individual three-dimensional trajectory of the BCOM while moving on a treadmill is a closed loop (Lissajous *contour*; Pratt et al., 2003; Reisman et al., 2007; Minetti, 2009) following the same pattern at each stride.

To describe this closed loop, we started to develop a method using Fourier analysis (Nessler et al., 2009; for more details, see chapter 6), truncated to the 6th harmonic, for each of the 3 spatial coordinates of the BCOM, with time as the independent variable. This leads to Lissajous *contours*, made up of 3 parametric equations which are characterised by 6 harmonic coefficients and 6 phases (for more details, see chapter 6).

Moreover, this graphical description (Lissajous *contour*) would clarify the most significant individual difference in anterior/posterior, vertical and medial/lateral displacement. These 6 harmonic coefficients and 6 phases defined the so-called *Digital Locomotory Signature* (DLS; Figure 3.1, 3.2 and 3.3. In these figures onwards, the right is on the left side and *vice versa*, according to the graphical requirements implemented in *Grapher* software. For more details, see chapter 7, par. 2.1).

This important parameter DLS was defined for each gender, age group, type of locomotion, speed and gradient (for more details, see chapter 7).

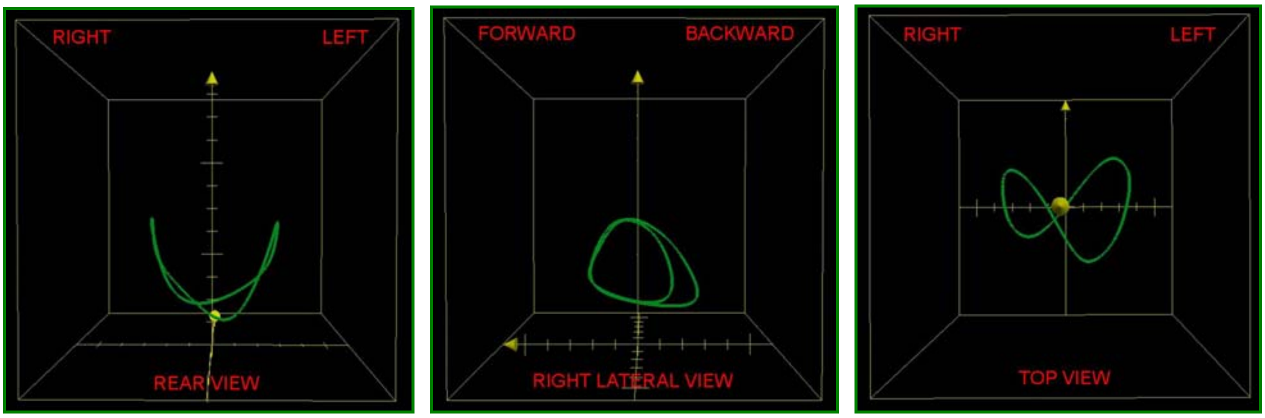

Figure 3.1. Average 3D *contours* in level walking at 1.39 m/s, males aged 25 to 35.

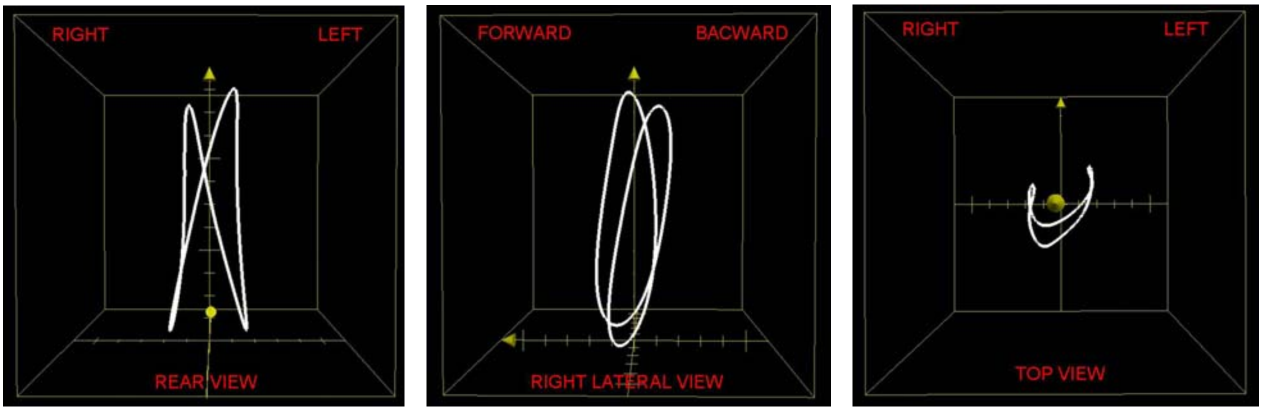

Figure 3.2. Average 3D *contours* in level running at 2.50 m/s, males aged 25 to 35.

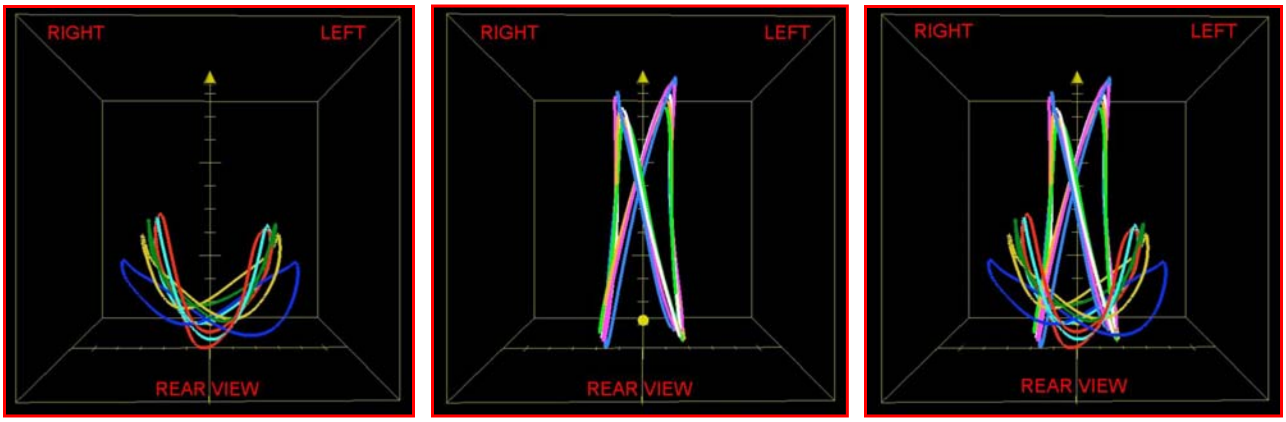

Figure 3.3. Average 3D *contours* in level walking (on the left) and running (in the middle) at all speeds (on the right), males aged 25 to 35.

Furthermore, we have calculated an important variable (the so-called *Symmetry Index*; for more details, see chapter 8) that contains and summarizes the most important information regarding right/left (a)symmetry in each movement direction (Kulagin et al., 1970; Draper, 2000; Sadeghi et al., 2000; Archer et al., 2006; Shorter et al., 2008; McFayden et al., 2009; Parkes et al., 2009; Starke et al., 2009). This index allowed us to fully complete the description of average 3D *contours* in each testing conditions.

Polar graphs (for more details, see chapter 9) allowed us to show, in a two-dimensional system, the combined pattern of both coefficient amplitude and phase.

Therefore, this mathematical and graphical method allowed us to characterize specific individual (a)symmetries in various normal conditions. When moving, humans show a tendency to turn in one direction (Souman et al., 2009). Therefore, we tried to verify which is the best turning direction in our studied people.

In fact, the final outcome was an initial comprehensive *database* of 'normality values' (reference equations/coefficients), describing normal locomotion in a set of different conditions (gender, age, type of locomotion, speed and gradient).

Consequently, it also becomes possible to extend the experiments to other conditions, and spread the advantages of this method to detect gait anomalies (Andriacchi et al., 1977; Cappozzo, 1984). Perspectives have included the ability to detect variations in locomotion dynamics such as those caused by training, passive aids, ageing, gait pathology and rehabilitation (Minetti, 2006). The quantitative evaluation of the changes in the global locomotion pattern, during and following rehabilitation treatments and training sessions, could help to modulate the interventions and to better understand their overall effects.

The knowledge and the development of this approach constitutes one of the most important and relevant theoretical and practical result of this study.

Furthermore, we have investigated and evaluated some important simple biomechanical variables, such as stride frequency, stride length and duty factor (for more details, see chapter 10). We have also calculated various complex biomechanical variables, such as mechanical external (see chapters 10 and 11) and i mechanical nternal work (see chapters 10 and 12), mechanical total work (see chapter 10) and energy recovery percentage (see chapters 10 and 11). Knowledge of these biomechanical variables is important both to extract and characterize the individual gait signature, and also to fully describe the mechanics of walking and running (Barrett et al., 2008; Chumanov et al., 2008; Kang et al., 2008; Hernandez et al., 2009; Røislien et al., 2009).

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PART 1

MATERIALS AND METHODS

Chapter 4

INSTRUMENTATION

1. INTRODUCTION

In this chapter, we are going to focus on the two different instruments we used in order to carry out all test experiments and protocols (see also chapter 5). For each piece of equipment, we have submitted: a) a brief review from the literature available to define and evaluate the main characteristics (also in terms of advantages and disadvantages); and b) a brief and simple presentation and illustration of the specific components and relative functions.

The equipment is illustrated in this order:

1. treadmill h/p/Cosmos (par. 2);
2. Vicon motion capture system (par. 3), to record kinematics data.

2. TREADMILL H/P/COSMOS

2.1. Treadmill: general characteristics and application

The treadmill could be defined as 'a piece of indoor sporting equipment primarily used to allow for the motions of walking or running while staying in one place' (Treadmill in Training - Wikipedia, the free encyclopedia, 2008).

Importantly, its main **advantages** could be summarized in: a) an increased endurance; b) a reduced impact; c) a reduced required calibration volume for capturing kinematic data (Schache et al., 2001); d) the possibility to enable exact calculation and adjustment of slope and speed; and e) by imposing a speed of movement, the mimic features of the natural constraint of an individual trying to get some place in a given amount of time could be observed (Bertram et al., 2001). However, its main **disadvantages** could be summarized in: a) an increased probability of personal injury (especially, at ankles and knees) if not used properly; and b) the lack of wind resistance makes locomotion on a treadmill easier than on an equal elevation gradient outdoors (Wheat et al., 2005).

The treadmill can be used in many situations and for many reasons: a) in research (e.g. to impose a stress or to simulate over-ground locomotion); b) in rehabilitation (e.g. to treat walking impairments; Harris-Love et al., 2001; Frenkel-Toledo et al., 2005; Chang et al., 2009); and c) in training (e.g. to improve aerobic fitness or burn calories, to perform physiological tests and to predict performance capabilities; Brouwer et al., 2009).

2.2. The belt system

The treadmill work principle is a belt system where the top of the belt moves to the rear so as to allow a subject to move along at an equal, and necessarily opposite, speed. It has been demonstrated that the belt speed variations during a stride highly depend on: a) the power of the treadmill used; b) the form of locomotion: walking or running; and c) the subject body mass (Savelberg et al., 1998).

2.3. The familiarisation period

Familiarisation may be defined as 'the process whereby differences in repeated measurements of a specific parameter stabilize to a certain level' (Lavcanska et al., 2005).

A familiarisation period for treadmill locomotion is required for a number of reasons: a) first of all, novice treadmill subjects often feel uncomfortable on being first exposed to the treadmill (Charteris et al., 1978; Wall et al., 1980; Schache et al., 2001; van de Putte et al., 2006); and b) secondly, the optical information of self-motion is different and needs to be reinterpreted, which logically takes some time.

In our test protocols (see chapters 5, par. 1.2, and 16, par. 2), a familiarisation period of at least 20 minutes was estimated for each subject, according to literature data.

2.4. Treadmill h/p/Cosmos Saturn 4.0

2.4.1. Dimensions

Treadmill walking and running tests (deeply described in chapters 5 and 16) were performed on a well-instrumented treadmill: h/p/Cosmos Saturn 4.0 (300/100r; Figure 4.1).

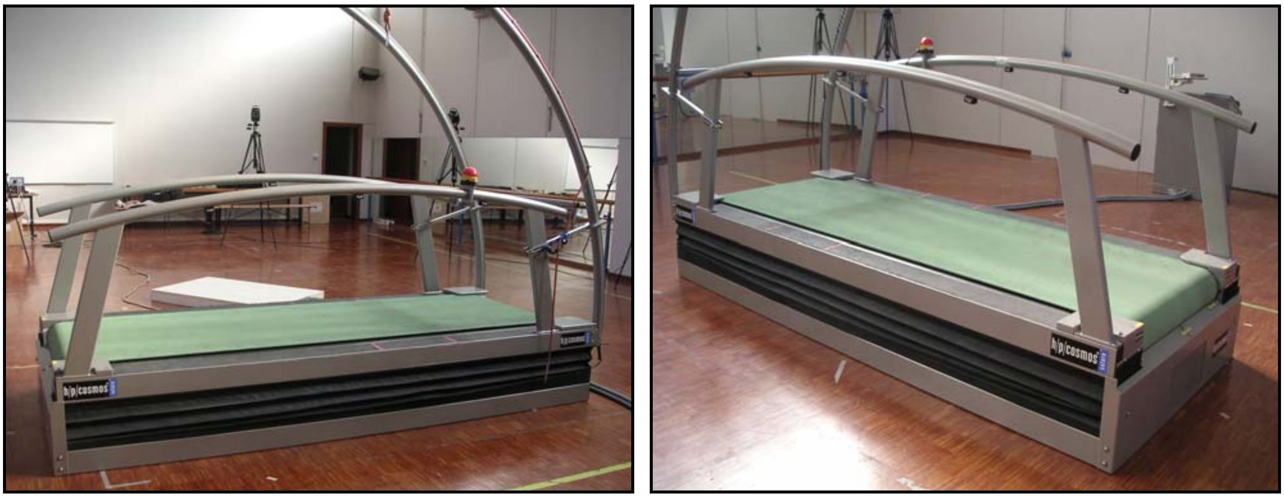


Figure 4.1. Treadmill h/p/Cosmos Saturn 300/100r (Cosmos, Germany).

Its main dimensions are: a) 300 cm in length; and b) 100 cm in width. These greater dimensions are based on the underlying ideas: 1) to isolate and study many repeated stride cycles; 2) to operate at a constant and stationary state (by means of controlled speed); and 3) to study types of locomotion occurring in medium-large volume (Minetti et al., 1993).

It also presents some handrails, collocated at the front and on the sides. Two curved anterior and superior bars accommodate a harness which can be attached to a rope for hoisting. This sling harness is usually used with children and older adults (two groups of people that present slight imbalance disorders), in order to prevent and avoid any falls. In our experiments this device was used when necessary. Finally, on the right handrail, there is a red button (Figure 4.2). When a subject is afraid or does not feel completely safe, he/she can press this button. As a result, the treadmill will immediately stop: this is a necessary safety measure (h/p/Cosmos Manual, 2005).

2.4.2. The display (configuration)

As shown below in Figure 4.2, keys contain all functions: a) start (or selected mode), stop and discontinue operation functions; b) alteration of elevation (i.e. down and up); c) modification of speed (i.e. acceleration and deceleration); and d) selection or confirmation of mode (manual, profile, cardio and test) of parameter (speed, time pulse …) and of options (measuring unit).

The mode *manual* (h/p/Cosmos Manual, 2005) was used in our experiments.

Figure 4.2. The keys and the display (on the left). Safety measure (on the right).

3. IMAGING MEASUREMENT TECHNIQUES: VICON MX SYSTEM

3.1. Introduction

Human movement analysis aims at gathering quantitative information about the mechanics of the muscular-skeletal system (e.g. during the execution of a motor task). In particular, information is sought concerning a) the movement of the BCOM and the relative movement and kinematics/kinetics of upper and lower limbs (Perry, 2005), b) the forces changing with the environment, and c) the resultant loads transmitted across sections of body segments or between body segments, or transmitted by individual body tissues such as muscles, tendons, ligaments and bones (Cappozzo et al., 2005).

The Chinese proverb *a picture is worth more than ten thousand words* (Winter, 2005) holds an important message for any human observer, including the biomechanics researcher interested in human movement. The only system that can possibly capture all the data is an imaging system because of the complexity of most movements (Lamoreux, 1971; Miller et al., 1973; Mansour et al., 1982; Girard et al., 1985; Belli et al., 1992; Vaughan et al., 1992; Aggarwal et al., 1999; Andriacchi et al., 2000; Sparrow, 2000; Wu et al., 2000; Moeslund et al., 2001; Hamilton et al., 2002; Wang et al., 2003; Robertson et al., 2004; Hoffman, 2005; Winter, 2005; Paroczai et al., 2006, Trewartha et al., 2008; ESMAC Hand Notes, 2009; Racic et al., 2009; Tulchin et al., 2009).

3.2. Different imaging techniques

Nowadays, in a laboratory, many different imaging techniques can co-exist. It becomes important to understand which is the better technique.

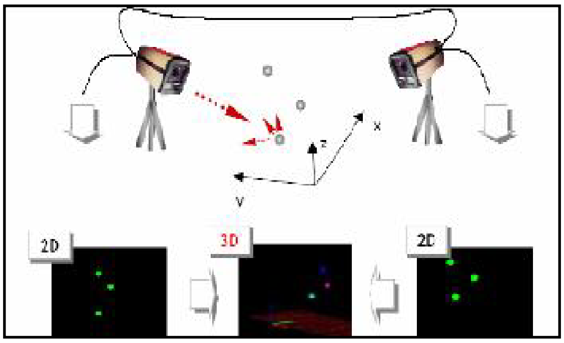
3.2.1. Photography

The development of photographic methods of recording a series of displacements during locomotion by the end of the 19th century encouraged researchers from different disciplines to study human motion (Vaughan et al., 1992; Aggarwal et al., 1999; Andriacchi et al., 2000). Experiments by the American photographer E. Muybridge when he photographed animals (e.g. horses trotting) and humans in motion (e.g. athletes doing a sport) perfected the study of animal and human locomotion (Muybridge, 1887a; 1887b; Miller et al., 1973; Mündermann et al., 2006; Richards, 2008; Racic et al., 2009). Later, E.J. Marey, W. Braune, O. Fischer and N. Berstein carried out experiments studying human locomotion (Marey, 1885; 1894; Mündermann et al., 2006; Richards, 2008; Racic et al., 2009).

Following these pioneers, lots of researchers (Saunders et al., 1953; Inman, 1966) from different disciplines studied human locomotion. Since the early seventies of the last century, biomechanics researchers used a technique similar to the techniques used by Marey for gait analysis and assessment.

3.2.2. Video recordings

Video recordings can be used to measure joint angles and velocities and to define gait signature from single or multiple cameras. This method has been added to the development of analysis software that greatly simplifies the analysis process and allows for analysis in 3 dimensions rather than 2 dimensions (Figure 4.3. 2D and 3D video-recordings; Wyss et al., 1981; Kuxhaus et al.,



2009). Video images have several potential advantages (see par. 3.3.2 below) over photography or radiography for the stereophotogrammetric reconstruction of landmark coordinates (Vaughan et al., 1992; Stevens, 1997). Nowadays, this imaging technique is the best and the most developed. Finally, in the following sections we will focus on it.

3.3. Modern motion capture systems

3.3.1. Introduction

Optoelectronic motion capturing is widely used in Biomechanics (Dillman, 1975; Bates et al., 1979; Krebs et al., 1985; Vagenas et al., 1992; Adrian et al., 1994; Novacheck, 1995; Hamilton et al., 2002; Archer et al., 2006; Windolf et al., 2008; Bevins et al., 2009; Hartog et al., 2009; O'Connor et al., 2009; Stokic et al., 2009).

Modern ***motion capture*** (or *mocap*) is a technique of digitally recording movements for entertainment, sport and medical applications (Cappozzo et al., 1975; Davis et al., 1991; Vaughan et al., 1992; Wu et al., 2000; Cicchella, 2002; Hartog et al., 2009). Instantaneous positions of markers located on the skin surface are then obtained using stereophotogrammetry (O'Connor et al., 2009), or *motion capture*, based optoelectronic sensors. Stereophotogrammetric methods are used to reconstruct 3D landmark coordinates from photography, radiography and video image (Robertson et al., 2004; Chiari et al., 2005). This theoretical background, deriving 3D coordinates of retro-reflective markers from several 2D camera projections, has been sufficiently established and extensively applied (Chen et al., 1994).

There are many types of imaging systems that can be used, each of them with specific properties (Vaughan et al., 1992; Robertson et al., 2004; Chiari et al., 2005). Since the 3D camera system is utilized in the analysis and evaluation of a subject's gait, it is important to evaluate the performance of each system (Ehara et al., 1997). Nowadays, gait analysis instrumentation is based on infra-red (IR) cameras. Computer-aided systems recording the 3D positions of the markers attached to the subject have been used in many laboratories to record gait cycles and produce patterns and plots/graphs to assess and hence diagnose (Girard et al., 1985).

3.3.2. The main characteristics of modern motion capture systems

In this section, we will present and discuss the main advantages and disadvantages of motion capture systems over traditional computer animation in a general 3D model. This knowledge contributes to give important details for biomechanics researchers, since these systems are more often developed and used in both research laboratory and sportive or clinical settings.

Motion capture systems offer several **advantages**: a) they are more rapid (e.g. even real time results) (Richards, 2008); b) all data are presented in an absolute spatial reference system, in a plane orthogonal to the optical axis of the camera (Winter, 2005); c) encumbrance to movement is minimal for most systems that use light-weight reflective markers and the time to apply the markers is minimal (Baker, 2006); d) the amount of work does not vary with the complexity or length of the performance; e) complex movement and realistic physical interactions such as secondary animation, weight and exchange of forces can be more easily recreated in a physically accurate manner; f) the operator can choose any camera angle desired from a scene, including angles that are difficult or impossible to film in live action situations (Kuxhaus et al., 2009); and g) motion capture technology allows one actor to play multiple roles within a single film.

Motion capture systems present simultaneously several **disadvantages**: a) specific hardware and special programs (such as software Workstation 5.1; see par. 3.4.3 below) are required to obtain and process the data; b) the capture system may have specific requirements for the space which it is operated in. Consequently, the results are limited to what can be performed within the capture volume without extra editing of the data; c) their accuracy usually cannot reach the standards of their active marker counterparts (Racic et al., 2009); d) the technology can become obsolete every few years; e) applying motion to quadrupeds can be difficult; f) movement does not follow the laws of physics can not be represented; g) if the computer model has different proportions from the capture subject, artefacts may occur; and h) marker properties, optical projections, video-digital conversion, camera configuration, lens distortion, calibration procedure influence the performance to various extents (Furnee, 1991).

3.3.3. Sources of inaccuracy affecting photogrammetric measurements

Instrumental errors are of two types: a) systematic (intrinsic and extrinsic; Wells et al., 1980; Schwartz et al., 2004; Alonso et al., 2005; Cereatti et al., 2006; Richards, 2008) and b) random.

a. In any case, the former type is associated with a model of the measurement system of limited validity, due either to photogrammetric calibration inaccuracies (see par. 3.4.3 below), or to non-linearities occurring during calibration. The magnitude of the systematic errors depends on the size of the measurement field and on the position that the marker assumes within it (Gazzani, 1993).

b. Random errors may be due to electronic noise and marker flickering (Chiari et al., 2005). In addition, inertial effects, skin deformation and sliding/displacement, which occur mainly in areas closer to the joints, represent an artefact which affects the estimation of the skeletal system kinematics.

As regards this research project, the optoelectric Vicon MX system (Vicon, Milan, Italy) was used. In the following section, we will describe its main components, characteristics and properties.

3.4. Vicon MX system

3.4.1. Introduction

The Vicon MX system is one of the most advanced optical motion capture systems available. Its 3D reproduction has many uses (Vicon Manual, 2002): a) medical assessment of movement disorders; b) understanding of athletic techniques; c) generating lifelike animation for movies and videogames; and d) incorporating motion into virtual environments for engineering design.

The Vicon system includes hardware and software applications for the complete control and analysis of motion capture (Vicon system in Motion capture techniques - Vicon, 2008).

To analyse and process the data, we only used the central application of the Vicon software, Workstation 5.1 (see par. 3.4.3 below).

3.4.2. Main components

Principal components are:

a) **eight infrared (IR) MX13** (1.3 million pixels) **cameras** (Figure 4.4), whose fundamental properties are summarized in Table 4.1 (Williams et al., 2009). Each camera consists of a video camera, a strobe head unit, optical filter and cable. Each MX13 camera is programmed with software/firmware to control its operation and enable it to perform its own onboard processing. Vicon MX automatically recognizes cameras and their relevant characteristics when they are plugged in (Vicon Manual, 2002). The cameras send out infrared light signals and detect the reflection from the markers placed on the body. Triangulation of the marker in space is possible because it is based on the angle and time delay between the original and reflected signal.

PERFORMANCE	MX13 CAMERA
Resolution (pixels)	1280 H x - 1024 V
Maximum Frame Rate (fps) at full resolution	482
Aspect Ratio	5:4
Sensor size (megapixels)	1:3
Sensor size (mm)	15.4 h x 12.3 v
Shuttered	Yes
Sensor Type	CMOS
VGA Monitor Mode	60 Khz h x - 50 Khz v

Table 4.1. Main properties of Vicon MX13 cameras.

Before a test, any cameras and their own parameter were properly adjusted (e.g. threshold, strobe, gain and circularity). In order to identify all camera parameters, it is sufficient to observe the calibration object in a number of different positions with respect to the cameras (Chiari et al., 2005). Each camera is mounted on a tripod (Figure 4.4), changeable in its dimensions (i.e. height and width). Furthermore, this device has to be collocated before acquiring data;

Figure 4.4. A Vicon MX13 camera and its tripod.

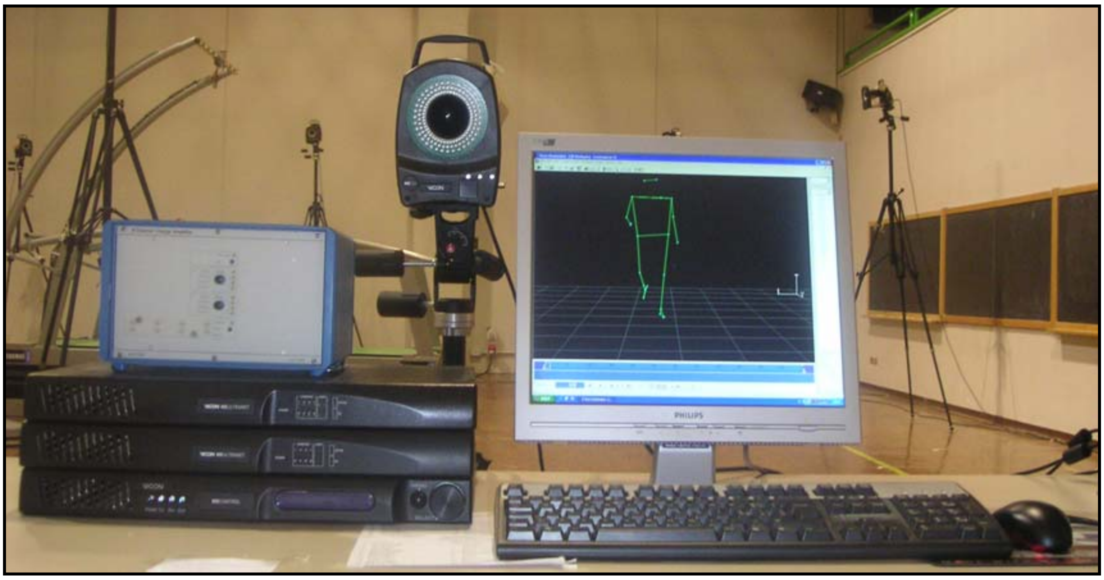


Figure 4.5. MXUltraNet and MXControl.

b) **MX Units** (Figure 4.5), made up of 2 MX Ultranet (to transfer information from each camera to computer unit) and of 1 MX Control (to integrate motion capture device to other analogue device, i.e. platforms; EMG systems; and so on);

c) **MX Calibration Kit** (see par. 3.4.4 below) and MX Accessory Kit;

d) **MX HOST PC**, where Vicon software and hardware (i.e. Workstation; BodyBuilder; Polygon; Plug-In-Gait; Plug-In-Modeller and Mobius) were plugged into.

In our experiments, the 8 MX13 cameras were positioned as illustrated in Figure 4.6.

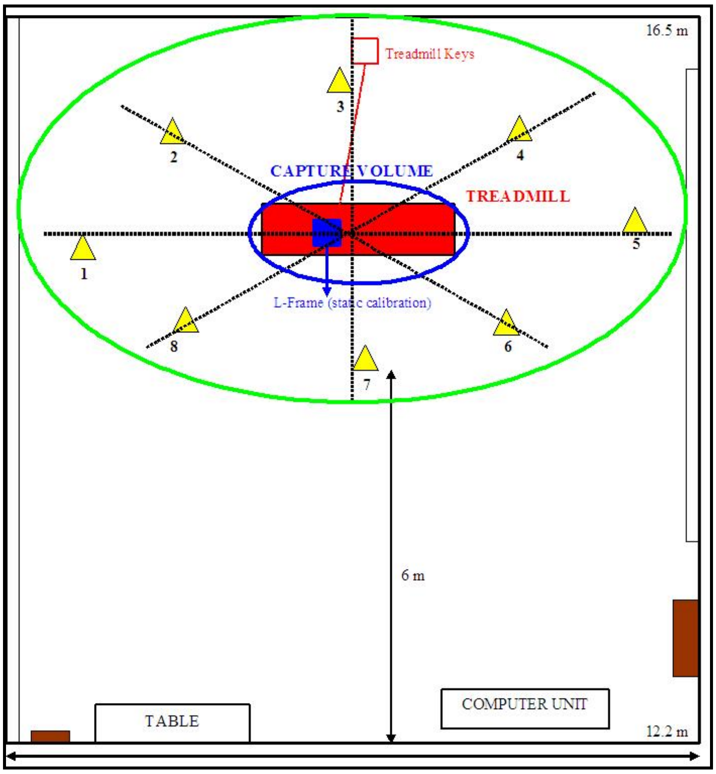


Figure 4.6. Setting of cameras in Biomechanics Laboratory of the Faculty of Exercise and Sport Science.

3.4.3. The Vicon software Workstation 5.1

Workstation is the central application of the Vicon software suited used to collect and process the raw video data (Vicon Manual, 2002; Davenport et al., 2009). It takes the two-dimensional data from each camera, combining it with calibration data (calibration parameters) and a user defined set of reconstruction parameters that depend on the volume and the type of capture, to reconstruct the equivalent digital motion in tri-dimensions (Figure 4.7).

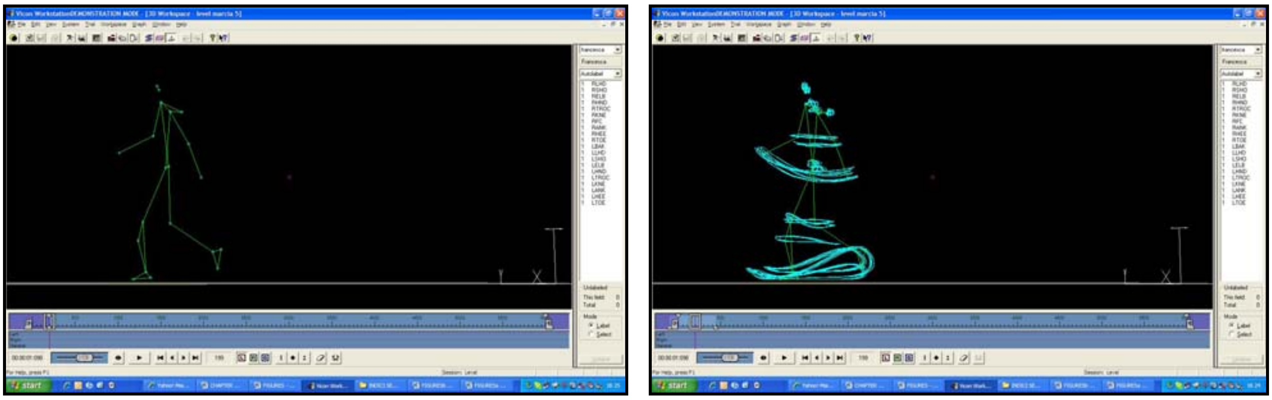


Figure 4.7. Stick diagrams (on the left) and marker trajectories (on the right), visualised by Workstation 5.1.

After this reconstruction, the data can be passed to other Vicon applications or software (see also chapter 6, par. 2) for analysis and manipulation.

3.4.4. Calibration

In general, the calibration of a motion capture system is one of the most important steps in capturing motion and recording data (Chiari et al., 2005; Richards, 2008). Indeed, calibration allows the system to define the capture volume (Vicon Manual, 2002; Robertson et al., 2004) and the relative position and orientation of the cameras. Furthermore, camera calibration aims at determining the geometric and optical characteristics of the cameras (internal parameters) and the position and orientation of the camera frame relative to the specific laboratory frame (external parameters), using 2D image points and the corresponding known 3D object points (Chiari et al., 2005; Baker, 2006). Once the capture volume has been defined, two types of calibration are exerted: 1) static and 2) dynamic.

1. Static calibration is used to set the origin and the direction of the axes by means of the L-frame (Figure 4.8, picture above. Calibration objects: L-Frame). Its duration is automatically defined by Vicon system. The L-frame was collocated according to standard three-dimensional coordinates defined by the ISB convention so that the laboratory (global) orthogonal coordinate system (frame) followed the left-hand rule (Wu et al., 1995; Racic et al., 2009). This biomechanical convention provides for (that): a) ***x-axis*** constitutes the direction of the movement progression so that the forward direction is positive and the backward direction negative; b) ***y-axis*** the medial-lateral direction so that the right side is positive and the left side negative; and c) ***z-axis*** the vertical direction of the movement, orientated vertically upward.

For more precise details about each axis-directions, see also chapter 6, par. 2.1.

2. Dynamic calibration is used to calculate the relative positions and orientations of the camera and to define the extrinsic parameters of the camera and the global coordinate system. We used the medium dynamic calibration object (called Wand; Figure 4.8, picture below. Calibration objects: Wand), length 240 mm with 3 markers (Ø = 14 mm). The duration is defined by the operator and it is dependent on the frequency acquisition; for our experiments, it is about 40.000 frame.

As far as our findings are concerned, all kinematic data was gathered using the 8 MX13 cameras sampling at 100 Hz (Bianchi et al., 1998; Chou et al., 2001; Preedy et al., 2001; Ivanenko et al., 2004; Baker, 2006; Mian et al., 2006; Brouwer et al., 2009; Mahaudens et al., 2009).

3.4.5. Marker set

Vicon MX is a passive marker system (Vicon Manual, 2002; ESMAC Hand Notes, 2009; Jordan et al., 2009). Passive reflective markers allow for very accurate measurement of movement using multiple cameras. These markers differ from active markers (Stevens, 1997; Mündermann et al., 2006; Grossman et al., 2008; Miana et al., 2009; Racic et al., 2009).

The position of markers are susceptible to two types of error, generally referred to as relative and absolute (Richards, 2008). They are often caused by movement of the soft tissue on which the markers are placed or by the subjective examiner technique (Moniz-Pereira et al., 2009).

The marker points need to be selected according to some experimental requirements: a) sufficient measurements should be available on the markers from the available cameras at any given time (Woltring, 1991; Karamanidis et al., 2003; Kuxhaus et al., 2009; Racic et al., 2009); b) the light emitted or reflected from markers should be oriented within the field of view of a sufficient number of cameras (Ehara et al., 1995); c) the relative movement between markers and underlying bone should be minimal (Cappozzo et al., 1996; Sati et al., 1996; Reinschmidt et al., 1997; Lucchetti et al., 1998; Mündermann et al., 2006; Groen et al., 2009); d) they should not be placed where they impede or block movement or where they are in danger of being knocked off; and e) it should be possible to place markers despite the presence of appliances such as orthoses, prostheses or external fracture fixators (Cappozzo et al., 1995).

In our experiments, the human body can be treated as a series of linked, rigid segments (see also chapter 1, par. 1.2.2). To be specific, 20 reflective markers (Ø = 14 mm) were placed on anatomical landmark points (Figure 4.9: left graph, and Table 4.2a). 18 of them were placed symmetrically (9 per each side; Koopman et al., 1995; Mian et al., 2006). 2 markers (RFC and LBAK) were placed asymmetrically because of Vicon system demands (Vicon Manual, 2002; Cereatti et al., 2006).

In this way, 12 body segments were defined (Figure 4.9: right graph, and Table 4.2b).

For more details, see also chapter 5, par. 2.1.

MARKER NAME ABBREVATION	COMPLETE MARKER NAME
RLHD	Right Lateral Head
RSHO	Right Shoulder
RELB	Right Elbow
RHND	Right Hand
RTROC	Right Trochanter
RKNE	Right Knee
RFC	Right Front Calf
RANK	Right Ankle
RHEE	Right Heel
RTOE	Right Toe
LBAK	Left Back
LLHD	Left Lateral Head

LSHO	Left Shoulder
LELB	Left Elbow
LHND	Left Hand
LTROC	Left Trochanter
LKNE	Left Knee
LANK	Left Ankle
LHEE	Left Heel
LTOE	Left Toe

Table 4.2a. Landmark points in the marker set of our experiments.

BODY SEGMENTS	**CORRESPONDING MARKERS**
Head	RLHD, LLHD
Trunk	RSHO, LSHO, LBAK
Right Upper Arm	RSHO, RELB
Left Upper Arm	LSHO, LELB
Right Fore Arm	RELB, RHND
Left Fore Arm	LELB, LHND
Right Thigh	RTROC, RKNE
Left Thigh	LTROC, LKNE
Right Shank	RKNE, RFC, RANK
Left Shank	LKNE, LANK
Right Foot	RANK, RHEE, RTOE
Left Foot	LANK, LHEE, LTOE

Table 4.2b. Body segments in the marker set of our experiments.

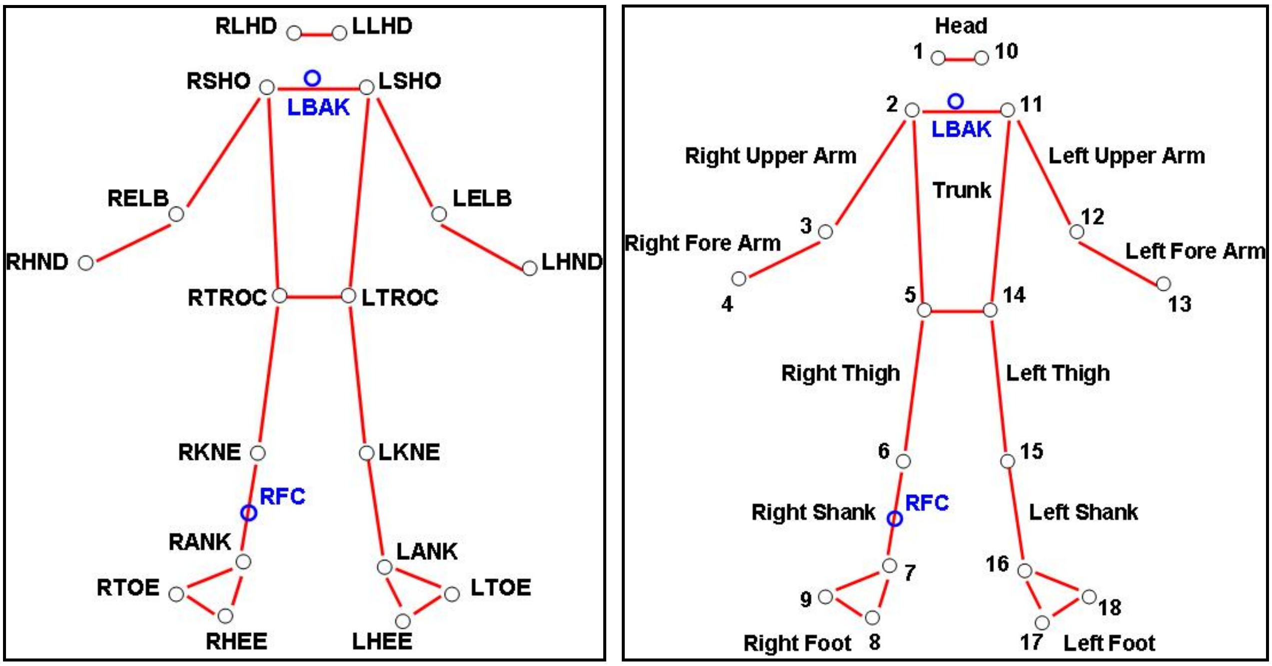


Figure 4.9. Marker set (on the left) and body segments (on the right).

In the most cases, the first trial a subject has to perform is the static condition (Figure 4.10a, at the end of the chapter). In fact, the subject has to stay in the centre of the capture volume, assuming a defined and stable upright attitude which has to be maintained for few seconds. While putting in the correct position the subject, the operator has to supervise (and control) that each marker will see simultaneously by, at least, two cameras. In this way, it becomes possible to choose the own marker set. In detail, each marker (contained in the apposite table on the right) has to be attach to its corresponding anatomical landmarks (McGinnis, 2005). Therefore, it is possible to copy all these information (using the proper function *Autolabel Calibration*, in Workstation Application) in order to paste them to all the following trials performed by the same subject. This automatic labelling avoids the manual intervention of the operator. It is particularly useful when a dynamic and animated performance has been expected. Clearly, it is necessary to check the automatic procedure in order: a) to verify the proper assignment of each marker; b) if necessary, to manually adjust (and correct) this nomination; and c) to discard some noises or reflections that appear in the capture volume (using the function *Delete Unlabed Trajectories*, in Workstation Application).

Once the static trial has been performed, it is possible to begin the recordings. The system will automatically labels each testing (static and dynamic) conditions (Baker, 2006; Richards, 2008).

At the end of the chapter, some stick diagrams and marker trajectories (from Workstation 5.1) of the marker set we used are proposed for the different testing conditions:

- level walking at 1.67 m/s, in a female aged 25 to 35 (Figure 4.10b and 4.10c);
- level running at 2.78 m/s, in the same female aged 25 to 35 (Figure 4.10d and 4.10e);
- uphill walking (25%) at 1.67 m/s, in the same female aged 25 to 35 (Figure 4.10f and 4.10g);
- downhill walking (-25%) at 1.67 m/s, in the same female aged 25 to 35 (Figure 4.10f and 4.10g).

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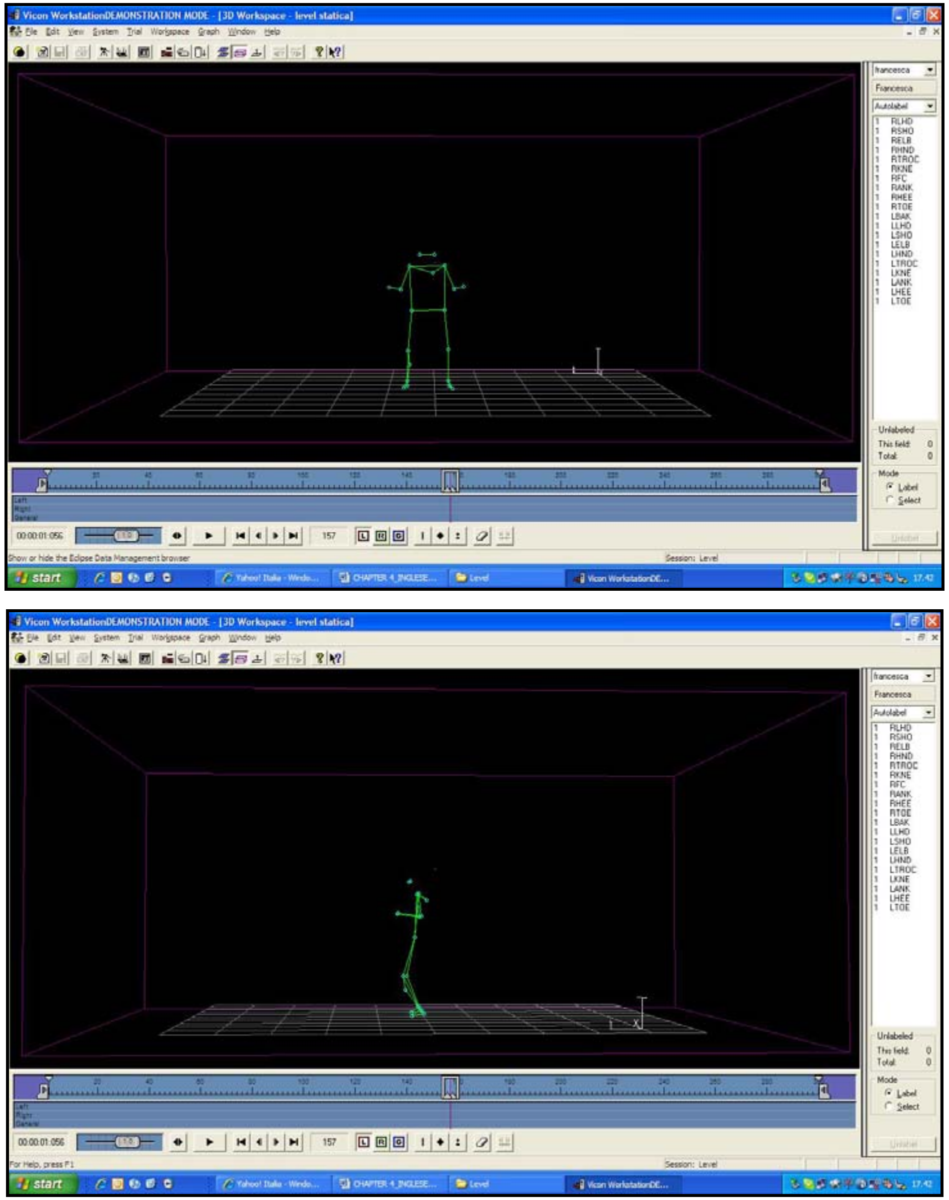

Figure 4.10a. Level static (front view, on the left; and left lateral view, on the right).

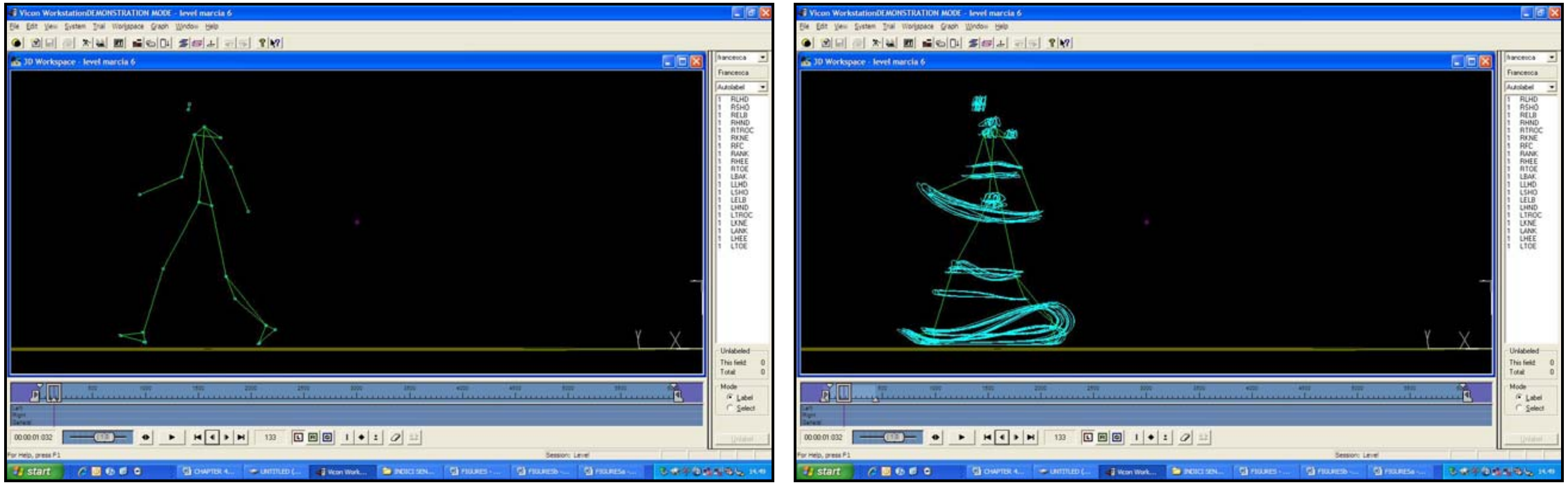

Figure 4.10b. Level walking at 1.67 m/s (left lateral view, on the left; and marker trajectories, on the right).

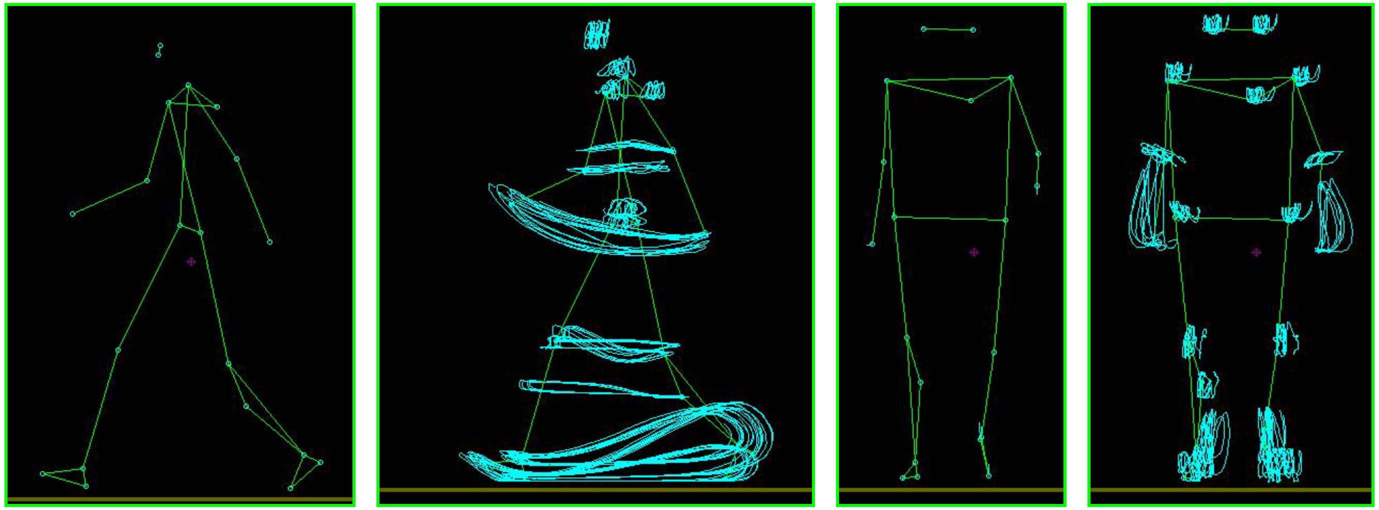

Figure 4.10c. Particulars of level walking at 1.67 m/s (left lateral view, on the left; and front view, on the right).

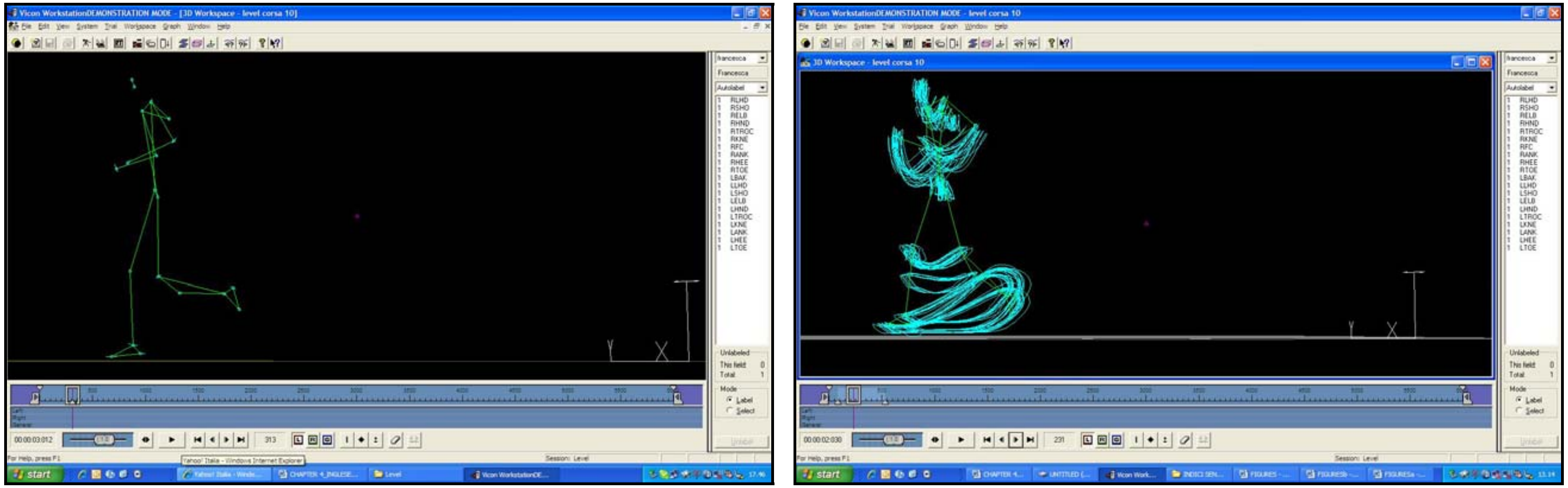

Figure 4.10d. Level running at 2.78 m/s (left lateral view, on the left; and marker trajectories, on the right).

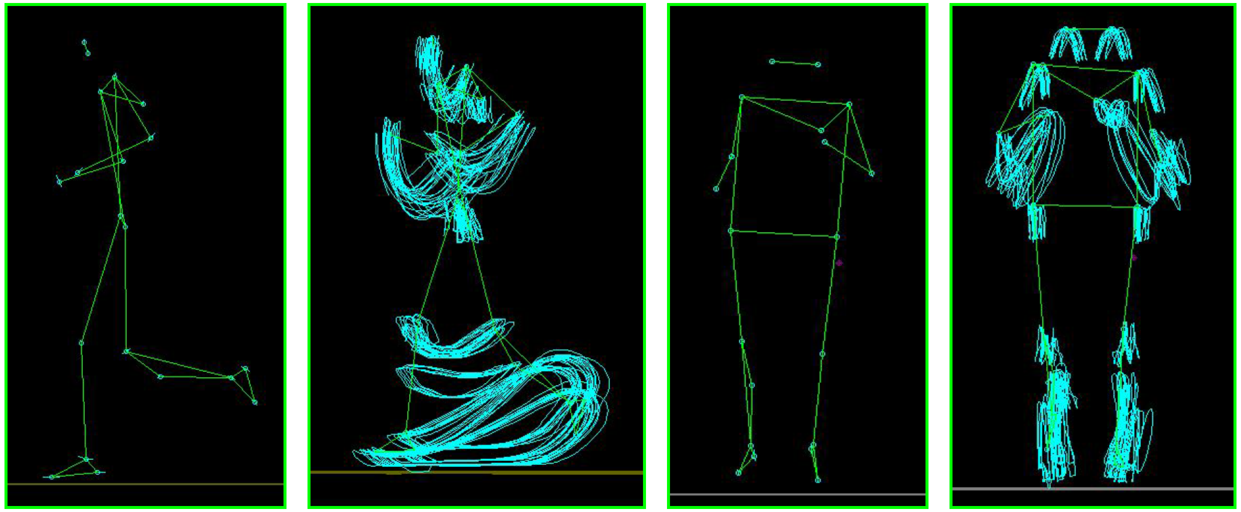

Figure 4.10e. Particulars of level running at 2.78 m/s (left lateral view, on the left; and front view, on the right).

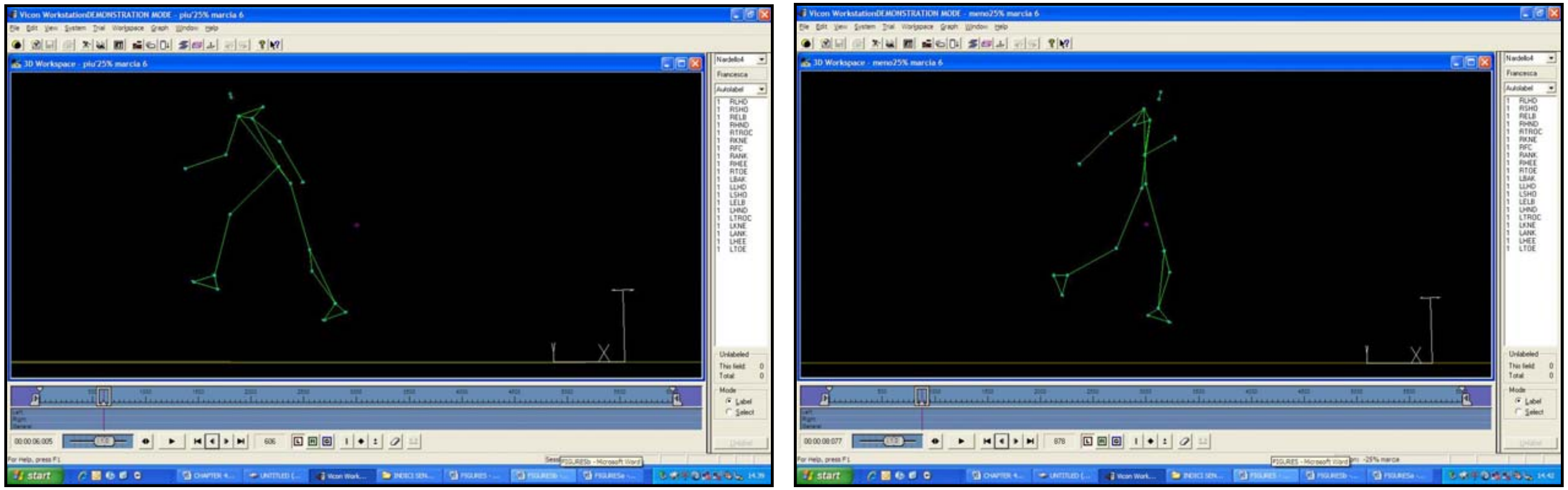

Figure 14.10f. Gradient uphill walking (25%) at 1.67 m/s (left lateral view, on the left) and gradient downhill walking (-25%) at 1.67 m/s (right lateral view, on the right).

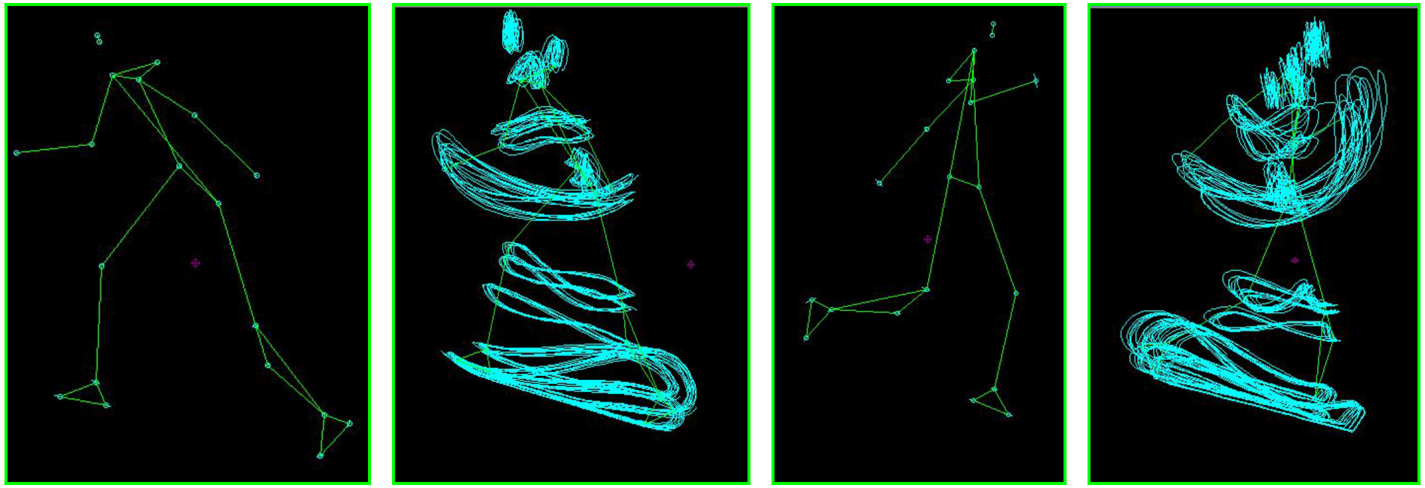

Figure 4.10g. Gradient uphill walking (25%) at 1.67 m/s (left lateral view, on the left) and gradient downhill walking (-25%) at 1.67 m/s (right lateral view, on the right).

Chapter 5

SUBJECTS

1. SUBJECTS

1.1. Introduction

We were interested in males and females of different genders and ages, who had to walk and run on a treadmill at different speeds and gradients. Consequently, we found and studied 70 healthy volunteers who participated to this study. Each of them was informed and gave his/her full consent prior to taking part (see Enclosed 5.1 below). These subjects came from a heterogeneous population of sedentary, normal or moderately active people. Only one important limitation to participation was imposed (the only criterion of inclusion): participants had to be unimpeded by neurological or musculoskeletal pathologies affecting gait (Danion et al., 2003; Whittington et al., 2009).

This sample didn't agree to statistics sample theory, because of the relevant novelty of the study (Armitage, 1975; Colton, 1979; Blailar III et al., 1988; Matthews et al., 1988; Glantz, 1994; Venables, 2002; Swinscow et al., 2004). According to this statistics theory, the minimum number of subjects to be examined is equivalent to:

$$t \bullet \frac{\alpha}{2} \bullet n-1 \bullet \frac{\mathrm{S.D.}}{\mathrm{Er\,min}} \quad \text{[Eq. 5.1]}$$

where $\frac{\alpha}{2}{}_{n-1}$ is the t-Student for the probability $\frac{\alpha}{2}$, at *n-1* degrees of freedom; *S.D.* is the standard deviation of *n* preliminary measurements (of the investigated parameter), carried out on the same subject, and at the same test conditions; and *Ermin* is the minimum error which can be tolerated.

1.2. How to catalogue subjects

These 70 subjects were then divided into several different groups, based on various conditions (Toro et al., 2007). First of all, the initial subdivision was based on gender. As a result, we studied:

- 35 male subjects;
- 35 female subjects.

Secondly, each of these new groups was further divided into seven subgroups based on age (ranging from 6 to 65). Therefore, we divided up the subjects into the following groups:

a) 5 males and 5 females aged 6 to 13;

b) 5 males and 5 females aged 14 to 17;

c) 5 males and 5 females aged 18 to 24;

d) 5 males and 5 females aged 25 to 35 (intermediate class);

e) 5 males and 5 females aged 36 to 45;

f) 5 males and 5 females aged 46 to 55;

g) 5 males and 5 females aged 56 to 65.

In this way, it was possible to study and investigate gait across ages ranging from 6 to 65. In the tables below (Table 5.1 and 5.2), each single subject age is presented.

MALES	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
#1	11	16	19	30	42	49	62
#2	10	16	22	34	40	46	63
#3	10	16	21	31	38	54	61
#4	12	16	20	26	37	51	57
#5	9	17	19	33	43	46	62

Table 5.1. Age of single males.

FEMALES	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
#1	13	14	24	27	39	48	64
#2	9	16	21	30	36	48	65
#3	11	17	20	27	42	54	57
#4	7	14	23	27	43	46	56
#5	9	17	24	34	44	55	59

Table 5.2. Age of single females.

During kinematic data analysis (see also chapter 6, par. 2), we have weighed up the effect of each age groups, aside from age of single males/females studied.

1.3. Anthropometric measurements

Body mass (kg) and height (cm) of each subject were recorded. The mean anthropometric measurements for any age group are then presented in the tables below (Table 5.3 and 5.4):

MALES	Body Mass (kg) ± S.D.	Height (cm) ± S.D.
6 - 13 y	37.4 ± 4.8	145.2 ± 6.0
14 - 17 y	61.6 ± 12.6	174.2 ± 10.9
18 - 24 y	69.7 ± 13.7	178.7 ± 6.6
25 - 35 y	82.2 ± 9.4	182.0 ± 4.6
36 - 45 y	84.6 ± 3.6	182.8 ± 3.4
46 - 55 y	92.6 ± 22.6	180.4 ± 8.1
56 - 65 y	88.6 ± 11.3	180.2 ± 1.8

Table 5.3. Average male anthropometric measurements.

FEMALES	Body Mass (kg) ± S.D.	Height (cm) ± S.D.
6 - 13 y	35.6 ± 8.7	140.2 ± 9.5
14 - 17 y	58.0 ± 5.8	165.6 ± 5.2
18 - 24 y	54.6 ± 3.8	165.0 ± 3.7
25 - 35 y	54.6 ± 6.1	169.6 ± 6.3
36 - 45 y	63.4 ± 15.9	169.6 ± 6.4
46 - 55 y	64.0 ± 10.2	163.6 ± 10.3
56 - 65 y	58.8 ± 6.9	161.0 ± 4.2

Table 5.4. Average female anthropometric measurements.

2. PROTOCOL TEST

2.1. General minimum protocol test

The final criterion of classification was based on the protocol test. First of all, markers were placed on the anatomical landmark points (Figure 5.1; see also chapter 4, par. 3.4.5).

Figure 5.1. Markers placed on a subject
(frontal view, on the left; right lateral view, in the middle; and posterior view, on the right).

Secondly, a static trial (Figure 5.2) was performed in order to create (and perform) the aforementioned *Autolabel Calibration* (see also chapter 4, par. 3.4.5). In our experiments, this condition took 3 seconds and it was carried out for each subject.

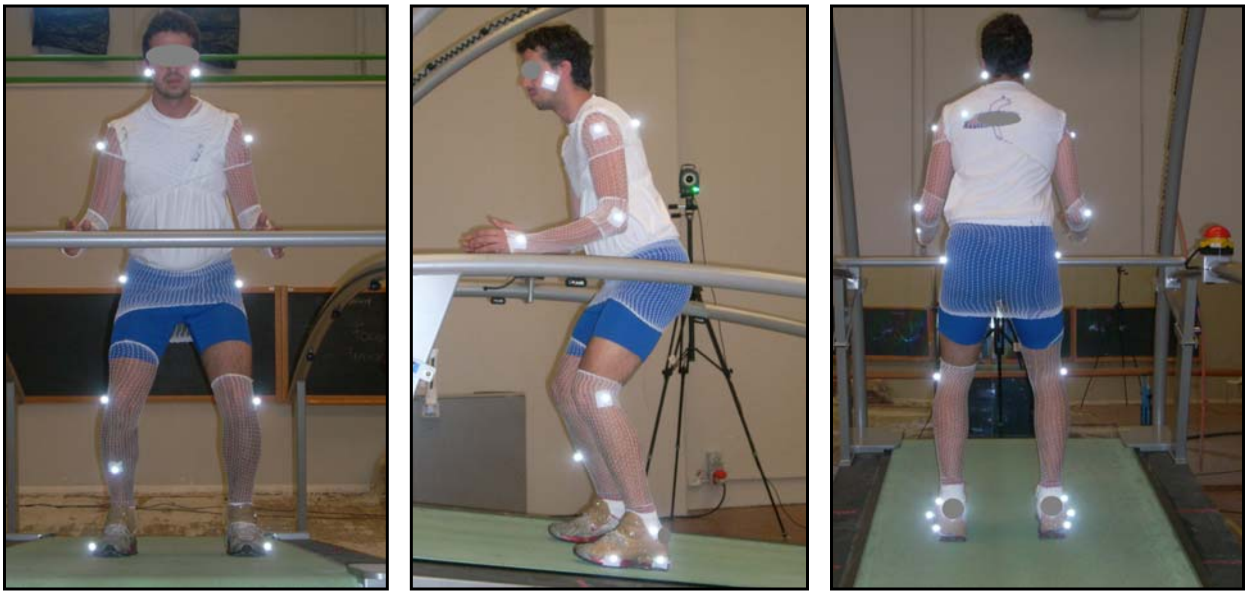

Figure 5.2. A static trial in our experiments
(frontal view, on the left; left lateral view, in the middle; and posterior view, on the right).

Finally, each subject had to perform a single test session (minimum protocol test), including:

1. a brief period of familiarisation on the treadmill (at least 20 minutes, according to the documentation data; see also chapter 4, par. 2.3. This time is primarily relevant for children and old adults; Griffin et al., 1999; Lavcanska et al., 2005; Ortega et al., 2005; Wass et al., 2005; Gottschall et al., 2006; Mian et al., 2006; Segers et al., 2006; Amorin et al., 2009; Vasudevan et al., 2009);
2. **level walking** at five different speeds (Hof et al., 2002; Bruijn et al., 2009; Chang et al., 2009): 0.83 m/s (= 3 km/h), 1.11 m/s (= 4 km/h), 1.39 m/s (= 5 km/h), 1.67 m/s (= 6 km/h) and 1.94 m/s (= 7 km/h) for walking (step 0.28 m/s = 1 km/h);
3. **level running** at five different speeds: 1.94 m/s (= 7 km/h), 2.22 m/s (= 8 km/h), 2.50 m/s (= 9 km/h), 2.78 m/s (= 10 km/h) and 3.06 m/s (= 11 km/h) for running (step 0.28 m/s = 1 km/h).

In order to avoid drawbacks, each test started at the lower speed. When a subject needed to, it was possible to pause and stop the test. On average, the test session took 1 hour and 30 minutes.

To sum up, each subject carried out 10 trials (700 conditions, in total).

There were some exceptions:

- two young females (aged 6 to 13) were not able to walk at the higher speed of 1.94 m/s;
- one young female (aged 6 to 13) was not able to run at 2.78 and 3.06 m/s;
- two young females (aged 6 to 13) were not able to run at 3.06 m/s;
- two adult females (aged 36 to 45 and 56 to 65) were not able to run at 3.06 m/s;

- three adult subjects (two females aged 46 to 55, and one male aged 56 to 65) didn’t perform running trials.

On the whole, **23 trials** were not performed.

During each test, the subject had to walk and run as naturally and regularly as possible. He/she also had to keep to in the middle of the treadmill, looking straight ahead.

Each speed was proposed in a random order: walking was also mixed with running, and *vice versa*. This was fundamental since postural and psychological adjustments could be discarded. Each testing condition (speed and gradient) was maintained for at least 60 seconds. This represents a time long enough to record an acceptable number of gait strides (Minetti et al., 1992; Gard et al., 2004). Finally, in order to avoid external influences on the individual pattern of walking and running, subjects were never aware when registration data began and/or stopped.

All testing was carried out utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

2.2. Specific protocol test

After the static trial (repeated in each session), males and females of the intermediate class (10 subjects altogether aged 25 to 35, in all; Kubo et al., 2003; Fell et al., 2008) also had to perform an additional specific protocol test. They all had to walk and run at the same speed, performed on the level condition (see par. 2.1 above), but on different gradients (Minetti et al., 1993; Grasso et al., 1998; Kang et al., 2002; Minetti et al., 2002).

Both for walking and running, the gradients involved in the study were:

- **uphill**: +5%; +10%; +15%; +20%; and +25%;
- **downhill**: -5%; -10%; -15%; -20%; and -25%.

In order to avoid drawbacks, each test started, on any gradient condition, at the lower speed.

On average, each subject had to perform six subsequent test sessions in order to complete all the protocol. Each session went on for usually 2 hours.

To sum up, 25 to 35 years males/females carried out 100 trials (<u>1100 conditions</u>, in total).

There were some exceptions:

- one female was not able to walk at 1.94 m/s, at +20% and +25%; to run at 2.78 and 3.06 m/s at +20%; and finally to run at all speeds, at the gradient of +25%;
- one female was not able to walk at all speeds, at +25%; to run at all speeds at +20% and +25%; and finally to run at 2.78 and 3.06 m/s, at -25%;
- moreover, one female was not able to walk at 1.94 m/s, at +15%, +20% and +25%; to run at all speeds at +10%, +15%, +20% and +25%;

- one male was not able to run at 3.06 m/s, at +25%;
- one male was not able to run at 2.78 and 3.06 m/s, at +20% and +25%.

On the whole, **54 trials** were not performed.

Trials at the higher speeds and gradients (e.g. walking at 1.67 and 1.94 m/s, at +20% and +25%; and running at 2.78 and 3.06 m/s, at +20% and +25%) were maintained for at least 30 seconds, according to literature data.

All testing was carried out utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

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Department of Neurological and Visual Sciences
PhD Program in Exercise and Human Movement Science

HUMAN LOCOMOTION: SYMMETRY IN THE 3D TRAJECTORY OF THE BODY CENTRE OF MASS

Thank you for your taking part in this scientific experiment. Before starting, you will be given some information about why the exam is being carried out.

The aim of this project is to describe kinematic variables of the centre of mass of the human body (BCOM) over time and space domains. We will develop both a mathematical method and a valid evaluation protocol to explain the three-dimensional displacement of the BCOM.

To reach this main goal, we will study healthy people (both males and females, ranging from 6 to 65) moving on a treadmill (walking and running, in particular). As a result, you have to perform some walking and running trials at different speeds (from 0.83 to 1.94 m/s, for walking; and from 1.94 to 3.06 m/s, for running; step 0.28 m/s) and gradients (level, uphill and downhill).

20 reflective markers will be placed on the anatomical landmark points. A motion capture system will record kinematic data, in order to describe the three-dimensional trajectory of the BCOM, and to characterize the individual specific Digital Locomotory Signature. Each speed and gradient condition will be maintained for at least 60 seconds. In order to avoid drawbacks, each test will start on level conditions, at the lower speed.

All testing will be performed utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

We assure you that all data will remain anonymous and privacy will be guaranteed. Furthermore, data will only be utilized as regard this scientific research project.

Verona, date ………………

Subject’s signature

………………………………………………………………

Researcher’s signature

…………………………………………………………………

Enclosed 5.1. Informed consent to participate in this study.

Dipartimento di Scienze Neurologiche e della Visione
Corso di Dottorato in Scienze dell'Esercizio Fisico e del Movimento Umano

LOCOMOZIONE UMANA: SIMMETRIA NELLA TRAIETTORIA TRI-DIMENSIONALE DEL CENTRO DI MASSA CORPOREO

RingraziandoLa per avere aderito a questa sperimentazione scientifica, La informiamo sulla natura delle valutazioni che effettueremo e relative motivazioni.

Obiettivo di questo studio sono la descrizione e la caratterizzazione, nello spazio e nel tempo, delle variabili cinematiche del centro di massa corporeo globale (CMC) in soggetti normali e patologici durante diverse forme di locomozione. A questo riguardo, saranno sviluppati a) un metodo matematico e b) un protocollo di valutazione della traiettoria tri-dimensionale di suddetta variabile.

Per potere raggiungere questo obiettivo, abbiamo la necessità di effettuare su di Lei prove di marcia e corsa su ergometro trasportatore variandone i parametri di 'velocità' (da 0.83 a 1.94 m/s per la marcia; da 1.94 a 3.06 m/s per la corsa; con uno step di 0.28 m/s) e 'pendenza' (piano, salita e discesa). Le saranno applicati, in maniera non invasiva e su principali punti di repere anatomici, marker riflettenti, i movimenti dei quali verranno registrati automaticamente e digitalmente da un sistema optoelettronico di analisi del movimento al fine di caratterizzare l'individuale e specifica Impronta Digitale Locomotoria.

La durata di ogni test sarà proporzionale alla Sua capacità di sostenere lo sforzo fisico richiestoLe alle diverse 'velocità' e 'pendenze' di locomozione. Per motivi di sicurezza, inizieremo con le prove in piano ed a basse velocità.

Le prove sperimentali verranno interamente effettuate presso il Laboratorio di Biomeccanica della Facoltà di Scienze Motorie. Le ricordiamo che tutti i dati raccolti sono strettamente coperti da privacy, utilizzati solo a scopo di ricerca scientifica e coperti da anonimato.

Verona, data ………………

Firma, per presa visione, del soggetto sottoposto alle valutazioni
……………………………………………………………

Firma del ricercatore
……………………………………………………………

Allegato 5.1. Consenso informato per partecipare allo studio.

Chapter 6

METHOD TO ANALYSE THE 3D DISPLACEMENT OF THE BODY CENTRE OF MASS

1. INTRODUCTION

Our hypothesis is that at every gait, speed and gradient, a unique 3D *contour* is associated, and differences are detectable between gender and across ages (see also chapter 3; Minetti, 2006). In order to develop both a mathematical method and a valid evaluation protocol to explain the three-dimensional displacement of the BCOM:

1. we analysed all kinematic data (see par. 2 below) to describe and obtain information about the three-dimensional displacement of the BCOM;
2. for the three-dimensional trajectory of the BCOM can be described as a closed loop, following the same pattern at each stride (Lissajous *contour*), we developed a method using Fourier analysis (see par. 3, 4 and 5 below) whose mathematical method allowed us to describe this closed loop.

In the following sections, each of these steps is briefly presented and analysed.

2. KINEMATIC DATA ANALYSIS

2.1. Introduction

We analysed and elaborated kinematic data by means of a custom-written LabVIEW software (Minetti et al., 1993; Ardigò et al., 2005). This *ad hoc* written software provides that: a) ***x-axis*** constitutes the direction of the movement progression so that the forward direction is positive and the backward direction negative; b) ***y-axis*** constitutes the medial-lateral direction so that the right side is negative and the left side positive; and c) ***z-axis*** constitutes the vertical direction of the movement, orientated vertically upward. As a consequence, the biomechanical convention described in chapter 4 (par. 3.4.4) has been modified (according to the right-hand rule).

However, in this software the following notation has been used: x for forward, y for vertical and z for lateral direction. Therefore, we have respected this convention, as well.

- Each kinematic test/data was saved in *.c3d format (software Workstation 5.1; Vicon Manual, 2002; see also chapter 4, par. 3.4.3). After this preliminary procedure, the following steps were carried out on a Macintosh notebook (APPLE, USA), using *Classic* Application.

- Firstly, each new file (*.c3d format) was automatically converted into a similar file *.txt format, by means of a public LabVIEW 6.1 *.vi, called *C3D Reader* (Figure 6.1, at the end of the chapter). This file contained the three-dimensional displacement (anterior/posterior, vertical and medial/lateral direction) for each marker. In this phase, it was already possible to choose only the displacement of the 18 significant markers (excluding asymmetrical ones, like RFC and LBAK; see also chapter 4, par. 3.4.5).
- The file *.txt format was then analysed by means of a LabVIEW 2.2.1 *.vi, called *ViconGraph* (see some examples in Figure 6.2 and 6.3, at the end of the chapter). This Application allowed us to check and examine each test by the function 'Movement: vertical direction (y) *versus* frame' (Figure 6.4, at the end of the chapter). As a result, this step allowed us to individualise and pick out (and note in an Excel file) both the *start* and the *end frame* for each trial. It was also possible to count strides (in order to define the correct *stride number*). These parameters are important to characterize some simple biomechanical variables, such as stride frequency (see also chapter 10, par. 2).
- The same file *.txt format was then analysed with another LabVIEW 6.1 *.vi, called *Extract 18 mrk* (Figure 6.5). Before this application started, we had to collocate the start and the end frame in the appropriate control. In this way, the *.vi extracted only the frame range previously chosen. A new file *.extr format was then created.
- Finally, this last file *.extr format was analysed by means of a LabVIEW 2.2.1 *.vi, called *Motion Analysis Filter* (Figure 6.6). In order to start this last application, the main anthropometric measurements of each subject had to put into the software: body mass (kg) and height (mm). We also had to choose the direction of locomotion (forward or backward), and to add the eventual gradient at which the trial would be performed.

Other relevant anthropometric parameters (Dempster et al., 1959; 1967; Contini, 1972; Miller et al., 1973; Winter, 1979; Jensen, 1986; Hinrichs, 1990; LeVeau, 1993; Gard et al., 2004; Robertson et al., 2004; Scheleihauf, 2004; Dumas et al., 2007; Grimshaw et al., 2007) were necessary.

They were: a) **mass (%)** of each body segment (Miller et al., 1973; LeVeau, 1993; Robertson et al., 2004; Winter, 2005; Grimshaw et al., 2007; Richards, 2008); b) **proximal location of BCOM** of each segment (Miller et al., 1973; LeVeau, 1993; Robertson et al., 2004; Winter, 2005; Grimshaw et al., 2007; Richards, 2008); and c) **radius gyration** (Reid et al., 1990; LeVeau, 1993; Zatsiorsky, 2002; Grimshaw et al., 2007; Richards, 2008).

For the anatomical definition of each body segment see also chapter 4, par. 3.4.5. Values of each parameter were taken from Winter's anthropometric tables (2005), for males (Table 6.1); and estimated from Zatsiorsky et al. (1990) and de Leva et al. (1996), for females (Table 6.2).

More information are contained in the enclosed CD (First Study, Chapter 6, Anthropometry).

SEGMENTS	Segment Definition	Mass (%)	Proximal BCOM/ Segment length	Radius gyration/ Segment length
(TRUNK - HEAD)/2	Greater Trochanter/ Glenohumeral Joint C7 - T1 and 1st Rib/ Ear Canal	0.2890	0.6600	0.5030
THIGH	Greater Trochanter/ Femoral Condyle	0.1000	0.4330	0.3230
SHANK	Femoral Condyle/ Lateral Malleolus	0.0465	0.4330	0.3020
FOOT	Lateral Malleolus/ Head Metatarsal II	0.0145	0.5000	0.4750
UPPER ARM	Glenohumeral Axis/ Elbow Axis	0.0280	0.4360	0.3220
FORE ARM	Elbow Axis/ Ulnar Styloid	0.0220	0.6820	0.4680

Table 6.1. Male anthropometric parameters.

SEGMENTS	Segment Definition	Mass (%)	Proximal BCOM/ Segment length	Radius gyration/ Segment length
(TRUNK - HEAD)/2	Greater Trochanter/ Glenohumeral Joint C7 - T1 and 1st Rib/ Ear Canal	0.2824	0.6107	0.4827
THIGH	Greater Trochanter/ Femoral Condyle	0.1044	0.3819	0.3623
SHANK	Femoral Condyle/ Medial Malleolus	0.0517	0.4288	0.3209
FOOT	Lateral Malleolus/ Head Metatarsal II	0.0137	0.4546	0.5526
UPPER ARM	Glenohumeral Axis/ Elbow Axis	0.0263	0.4346	0.3141
FORE ARM	Elbow Axis/ Ulnar Styloid	0.0191	0.6798	0.4426

Table 6.2. Female anthropometric parameters.

Modalities *.bcm* and *Filter* were selected. Kinematics data was then low-pass filtered using a 'non adaptive' 5th order Butterworth filter with a 8.5 Hz cut-off frequency. This filter has been used because of previous experiences with unfiltered spatial data manually digitized on analogue movie-frames (Minetti et al., 1993; 1994) and it seemed to work well if compared both to no-filter and first-order filter conditions (Table 6.3). Finally, the Butterworth filter needed that 5 frames were added up to the file *.extr, too, by typing the proper value in the specific control of *Extract 18 mrk*.

Indeed, values of the speed (e.g. Vicon speed) derived by using this filter well concur with the imposed speed. Furthermore, the main biomechanical variables (i.e. energy recovery percentage, duty factor and mechanical external work) wholly concur with literature, as well. Therefore, these last settings were used in other previous experiments (Minetti et al., 1993; 1994).

Few comparisons obtained by applying different filter orders (Ferrigno et al., 1990) are contained in the enclosed CD (First Study, Chapter 6, Different filter orders: all stride cycles and limited number of strides).

SUBJECT	GAIT	FILTER	Vicon Speed (m/s)	Stride Frequency (Hz)	% Energy Recovery	Duty Factor (%)	W_{ext} (J/(kg·m))
1 male aged 25 to 35	level walking 0.83 m/s	**NO FILTER**	0.872	0.8029	34.19	61.90	0.755
	level walking 1.11 m/s		1.100	0.8770	39.69	59.90	0.777
	level walking 1.39 m/s		1.410	0.9742	46.29	57.50	0.677
	level walking 1.67 m/s		1.673	1.0175	47.31	56.60	0.770
	level walking 1.94 m/s		1.901	1.0454	64.27	55.50	0.894

Table 6.3a. Biomechanical variable values analysing data with no filter.

SUBJECT	GAIT	FILTER	Vicon Speed (m/s)	Stride Frequency (Hz)	% Energy Recovery	Duty Factor (%)	W_{ext} (J/(kg·m))
1 male aged 25 to 35	level walking 0.83 m/s	**FILTER ORDER 1**	0.872	0.8029	47.34	61.90	0.412
	level walking 1.11 m/s		1.100	0.8770	56.18	59.70	0.399
	level walking 1.39 m/s		1.411	0.9742	63.09	57.50	0.361
	level walking 1.67 m/s		1.673	1.0175	64.27	56.50	0.405
	level walking 1.94 m/s		1.901	1.0454	60.80	55.50	0.472

Table 6.3b. Biomechanical variable values analysing data with filter order 1.

SUBJECT	GAIT	FILTER	Vicon Speed (m/s)	Stride Frequency (Hz)	% Energy Recovery	Duty Factor (%)	W_{ext} (J/(kg·m))
1 male aged 25 to 35	level walking 0.83 m/s	**FILTER ORDER 5**	0.874	0.8029	51.41	61.30	0.374
	level walking 1.11 m/s		1.102	0.8770	58.12	59.40	0.383
	level walking 1.39 m/s		1.413	0.9742	63.88	56.40	0.365
	level walking 1.67 m/s		1.676	1.0175	64.50	54.70	0.409
	level walking 1.94 m/s		1.905	1.0454	61.23	52.70	0.466

Table 6.3c. Biomechanical variable values analysing data with filter order 5.

This end step provided the file *.bcm format, containing the so-called three-dimensional displacement of the BCOM (see par. 2.2 below).

Single files *.bcm format in all age groups and gaits (males and females) are contained in the enclosed CD (First Study, Chapter 6, File *.BCM: files are catalogued as a function of age). Moreover, single templates we have written are contained in the enclosed CD (First Study, Chapter 6, Spreadsheet Motion Analysis Filter).

1. Firstly, *Motion Analysis Filter* also automatically provided us with trial speed (in km/h). The slight differences between speeds derived by comparing the measured *Motion Analysis Filter* speed (obtained by Vicon system, and measured during the contact period on the treadmill) to the imposed treadmill speed are probably due to the weight exerted by the subject who moves on the treadmill. Importantly, the Vicon speed estimation seems to be more accurate and precise because it takes into account a measurement related to the single subject (i.e. his/her mass and height). Herein, average values of speed obtained from the application *Motion Analysis Filter* for each age group (males: Table 6.4a, and females: Table 6.4b) are presented:

AGE GROUPS	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
LEVEL WALKING: AVERAGE SPEED (m/s), in MALES							
0.83 m/s (= 3 km/h)	0.848	0.838	0.834	0.848	0.823	0.833	0.843
1.11 m/s (= 4 km/h)	1.128	1.126	1.113	1.098	1.132	1.126	1.105
1.39 m/s (= 5 km/h)	1.385	1.400	1.377	1.392	1.402	1.386	1.389
1.67 m/s (= 6 km/h)	1.660	1.689	1.703	1.679	1.685	1.683	1.682
1.94 m/s (= 7 km/h)	1.916	1.969	1.969	1.943	1.961	1.975	1.951
LEVEL RUNNING: AVERAGE SPEED (m/s), in MALES							
1.94 m/s (= 7 km/h)	1.887	1.934	1.921	1.957	1.911	1.962	1.958
2.22 m/s (= 8 km/h)	2.159	2.228	2.215	2.217	2.217	2.234	2.235
2.50 m/s (= 9 km/h)	2.429	2.504	2.499	2.497	2.513	2.500	2.529
2.78 m/s (= 10 km/h)	2.711	2.792	2.776	2.794	2.817	2.780	2.766
3.06 m/s (= 11 km/h)	3.007	3.060	3.059	3.084	3.074	3.046	3.049

Table 6.4a. Average speed values in males of all age groups.

AGE GROUPS	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
LEVEL WALKING: AVERAGE SPEED (m/s), in FEMALES							
0.83 m/s (= 3 km/h)	0.840	0.832	0.833	0.846	0.834	0.856	0.847
1.11 m/s (= 4 km/h)	1.105	1.133	1.113	1.117	1.118	1.114	1.109
1.39 m/s (= 5 km/h)	1.377	1.397	1.383	1.389	1.403	1.379	1.389
1.67 m/s (= 6 km/h)	1.658	1.698	1.672	1.668	1.697	1.682	1.672
1.94 m/s (= 7 km/h)	1.943	1.960	1.920	1.931	1.938	1.949	1.971
LEVEL RUNNING: AVERAGE SPEED (m/s), in FEMALES							
1.94 m/s (= 7 km/h)	1.921	1.972	1.969	1.959	1.930	1.961	1.952
2.22 m/s (= 8 km/h)	2.219	2.233	2.199	2.133	2.226	2.249	2.200
2.50 m/s (= 9 km/h)	2.454	2.525	2.469	2.419	2.503	2.515	2.472
2.78 m/s (= 10 km/h)	2.725	2.782	2.742	2.797	2.761	2.797	2.772
3.06 m/s (= 11 km/h)	3.043	3.083	2.995	3.085	3.054	3.075	3.059

Table 6.4b. Average speed values in females of all age groups.

Consequently, the speed is graphically represented using these values: see chapter 8 onwards.

Importantly, we have decided not to use the **Froude Number** (Alexander et al., 1983; Bastien et al., 2003; Saibene et al., 2003; Alexander, 2004; Vaughan et al., 2005; Bullimore et al., 2006; Delattre et al., 2008; Usherwood et al., 2008; Delattre et al., 2009; Ruckstuhl et al., 2009; Starke et al., 2009) as the independent variable which normalizes walking speeds. Indeed, by comparing all age groups (males and females) in different testing conditions, slight significant differences have been found as a function of age only in young children (aged 6 to 13). These differences (i.e. a higher value of Froude Number) are probably due to their lower anthropometric dimensions. Moreover, it seems that the highest walking speed seem to be too fast for these age groups. Consequently, because of the poor significance of such results, we have taken the moving *Motion Analysis Filter* speed as the independent variable.

Otherwise, single values of Froude Number are contained in the enclosed CD (First Study, Chapter 6, Froude Number in level and gradient gaits - both males and females -).

2. Secondly, this application automatically gave us with some **simple** (stride frequency and duty factor or contact time) and **complex biomechanical variables** (mechanical external, internal and total work, mechanical vertical and forward work, and energy recovery percentage), referring to the trial being studied. All these significant biomechanical variables will be discussed in chapter 10.

3. Finally, it could visualize the main pattern of each energy (kinetic, potential and total energies) as a function of time (Figure 6.7). These measurements were also illustrated and discussed in chapter 11.

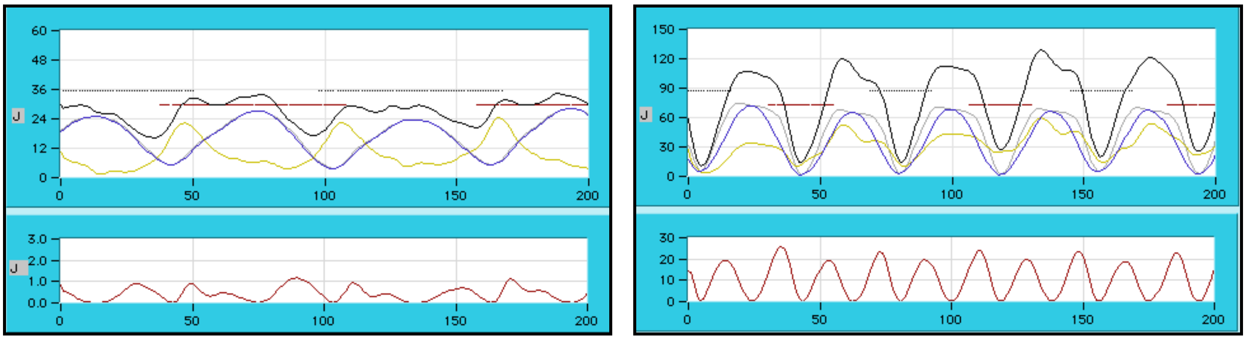


Figure 6.7. Patterns of horizontal kinetic (yellow), vertical kinetic (red), potential (blue) and total (black) energy as a function of time, in level walking at 1.11 m/s (on the left) and level running at 2.78 m/s (on the right), in a male aged 46 to 55.

- The file *.bcm format was then analysed by means of a LabVIEW 7.1 *.vi, called *Lissajous-Fourier BCOM Trajectory* (Figure 6.8, at the end of the chapter). This application is based on Fourier Analysis (see par. 4 and 5 below). *Threshold SD frequency and volume* was imposed 2. Modality *Exclude* was chosen/activated so that stride cycles were automatically picked out (Figure 6.9a).

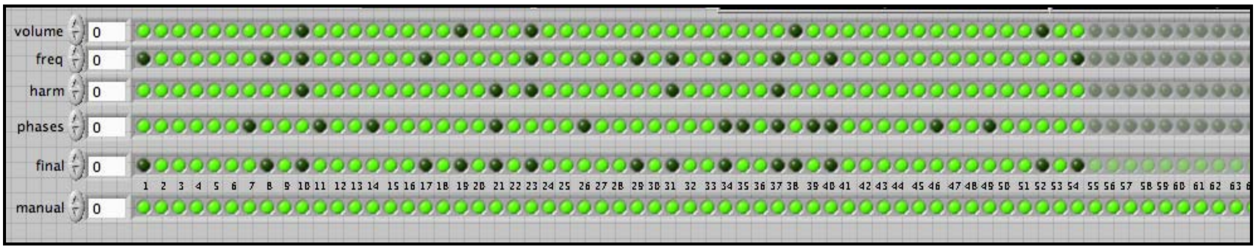


Figure 6.9a. Automatic selection of stride cycle in *Lissajous-Fourier BCOM Trajectory* *.vi.

It was possible to visualize the pattern of the BCOM in each plane, too (Figure 6.9b; Winter, 2005).

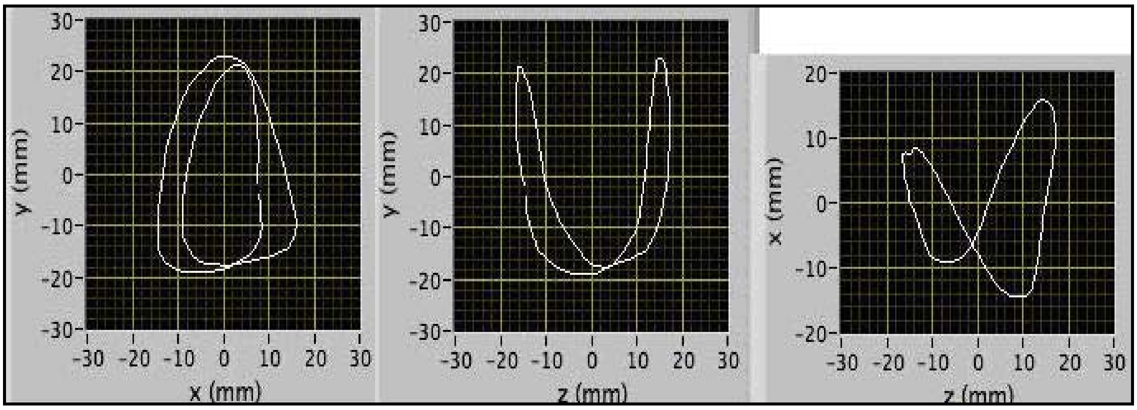
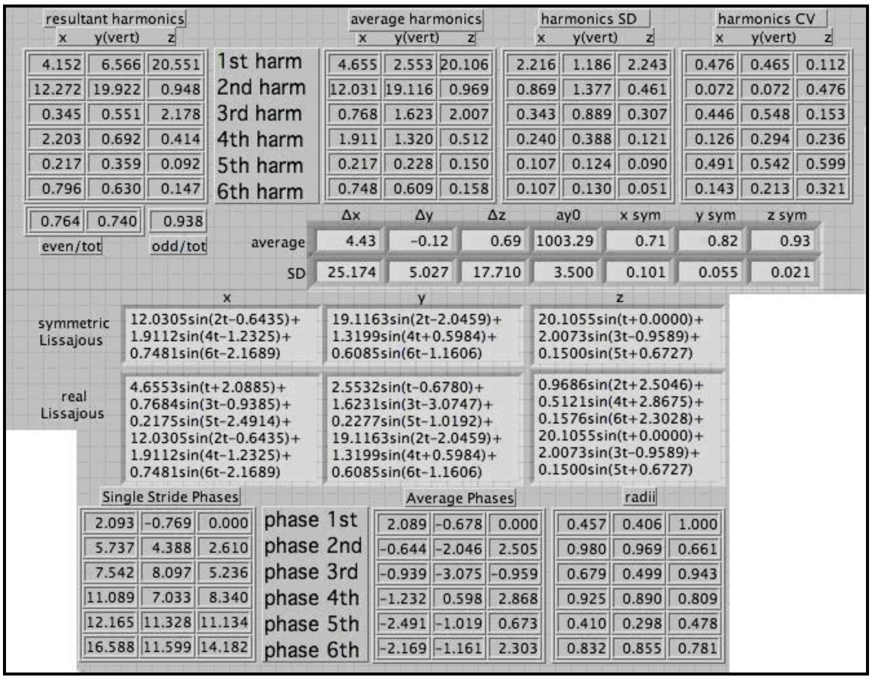


Figure 6.9b. Patterns of the BCOM in each plane in *Lissajous-Fourier BCOM Trajectory* *.vi.

The final result is a new file *.res format, mainly containing 6 **harmonic amplitudes** and 6 **phases** for each of the 3 spatial coordinates of the BCOM (Figure 6.10), with time as the independent variable. For more details, see par. 4 and 5 below.

Single files *.res format in all age groups and gaits (males and females) are contained in the enclosed CD (First Study, Chapter 6, File *.RES: files are catalogued as a function of age). Furthermore, the general template we have used is contained in the enclosed CD (First Study, Chapter 6, Motion Analysis Filter Res).



resultant harmonics				average harmonics			harmonics SD			harmonics CV		
x	y(vert)	z		x	y(vert)	z	x	y(vert)	z	x	y(vert)	z
4.152	6.566	20.551	1st harm	4.655	2.553	20.106	2.216	1.186	2.243	0.476	0.465	0.112
12.272	19.922	0.948	2nd harm	12.031	19.116	0.969	0.869	1.377	0.461	0.072	0.072	0.476
0.345	0.551	2.178	3rd harm	0.768	1.623	2.007	0.343	0.889	0.307	0.446	0.548	0.153
2.203	0.692	0.414	4th harm	1.911	1.320	0.512	0.240	0.388	0.121	0.126	0.294	0.236
0.217	0.359	0.092	5th harm	0.217	0.228	0.150	0.107	0.124	0.090	0.491	0.542	0.599
0.796	0.630	0.147	6th harm	0.748	0.609	0.158	0.107	0.130	0.051	0.143	0.213	0.321

even/tot		odd/tot
0.764	0.740	0.938

	Δx	Δy	Δz	ay0	x sym	y sym	z sym
average	4.43	-0.12	0.69	1003.29	0.71	0.82	0.93
SD	25.174	5.027	17.710	3.500	0.101	0.055	0.021

	x	y	z
symmetric Lissajous	12.0305sin(2t-0.6435)+ 1.9112sin(4t-1.2325)+ 0.7481sin(6t-2.1689)	19.1163sin(2t-2.0459)+ 1.3199sin(4t+0.5984)+ 0.6085sin(6t-1.1606)	20.1055sin(t+0.0000)+ 2.0073sin(3t-0.9589)+ 0.1500sin(5t+0.6727)
real Lissajous	4.6553sin(t+2.0885)+ 0.7684sin(3t-0.9385)+ 0.2175sin(5t-2.4914)+ 12.0305sin(2t-0.6435)+ 1.9112sin(4t-1.2325)+ 0.7481sin(6t-2.1689)	2.5532sin(t-0.6780)+ 1.6231sin(3t-3.0747)+ 0.2277sin(5t-1.0192)+ 19.1163sin(2t-2.0459)+ 1.3199sin(4t+0.5984)+ 0.6085sin(6t-1.1606)	0.9686sin(2t+2.5046)+ 0.5121sin(4t+2.8675)+ 0.1576sin(6t+2.3028)+ 20.1055sin(t+0.0000)+ 2.0073sin(3t-0.9589)+ 0.1500sin(5t+0.6727)

Single Stride Phases				Average Phases			radii		
2.093	-0.769	0.000	phase 1st	2.089	-0.678	0.000	0.457	0.406	1.000
5.737	4.388	2.610	phase 2nd	-0.644	-2.046	2.505	0.980	0.969	0.661
7.542	8.097	5.236	phase 3rd	-0.939	-3.075	-0.959	0.679	0.499	0.943
11.089	7.033	8.340	phase 4th	-1.232	0.598	2.868	0.925	0.890	0.809
12.165	11.328	11.134	phase 5th	-2.491	-1.019	0.673	0.410	0.298	0.478
16.588	11.599	14.182	phase 6th	-2.169	-1.161	2.303	0.832	0.855	0.781

Figure 6.10. An example of file *.res format containing harmonic coefficients and phases for each spatial coordinates (level walking at 1.39 m/s, in a male aged 25 to 35).

File *.res format also contained other important parameters such as:

- **n**. The first number on the left represents the total number of stride cycles performed during the trial; the middle number the stride cycles that the program have to reject; the last number on the right is automatically considered equals to 0;
- **range volume** in each movement direction (in mm). Its single values in all age groups and gaits (males and females) are contained in the enclosed CD (First Study, Chapter 6, Range Volume: level gait and gradient gait);
- **stride frequency** (± S.D.), in Hz. For more details concerning this biomechanical variable, see also chapters 10, par. 2 and 11, par. 1.2;
- **t1** (± S.D.) and **t2** (± S.D.), in sec. They represent the average time period in the harmonic's function;
- **Δx** (± S.D.), **Δy** (± S.D.) and **Δz** (± S.D.), in mm. They constitute the absolute average of the Δ displacement along each movement direction;
- **Ay0** (± S.D.), in mm. It represents the average vertical position of the BCOM. Single values of vertical position of the BCOM in all age groups and gaits (males and females) are contained in the enclosed CD (First Study, Chapter 6, Ay0: level gait and gradient gait);
- **x symmetrical** (± S.D.), **y symmetrical** (± S.D.) and **z symmetrical** (± S.D.), in mm. They represent the average displacement of the symmetrical coefficients along each movement direction;
- **CV** representing the coefficient of variation in amplitudes;
- **radii** (or *r*) for each direction, calculated as $\sqrt{\cos(\text{phase})^2 + \sin(\text{phase})^2}$. For more details concerning this circular variable, see chapter 9, par. 3.3;
- **Froude number** (see above);
- **cos (phase)** and **sin (phase)**. These measurements are used to calculate phase coefficient φ as arctan.2·(cos(phase); sin(phase)). In detail, since cos φ decreases from 1 to -1, if φ increases from 0° to 180°, φ is uniquely determined within this interval (Batschelet, 1981):

$$\varphi = \arccos x \ \text{ if } (-1 \le x \le 1,\ 0° \le \varphi \le 180°) \text{ [Eq. 6.1a]}$$

$$\varphi = \cos^{-1} x \ \text{ [Eq. 6.1b]}$$

Similarly, sin φ decreases from -1 to 1, if φ increases from -90° to 90° (Batschelet, 1981):

$$\varphi = \arcsin y \ \text{ if } (-1 \le y \le 1,\ -90° \le \varphi \le 90°) \text{ [Eq. 6.2a]}$$

$$\varphi = \sin^{-1} y \quad \text{[Eq. 6.2b]}$$

Furthermore, u = tan φ can be solved for φ uniquely, if φ is limited to the interval from -90° to 90° (Batschelet, 1981):

$$\varphi = \arctan u \ \ \text{if } (-90° < \varphi < 90°) \ \text{[Eq. 6.3a]}$$

$$\varphi = \tan^{-1} u \quad \text{[Eq. 6.3b]}$$

Finally, this application gives a first representation of the polar-log graph summarizing information of both harmonics coefficients and phases (Figure 6.11).

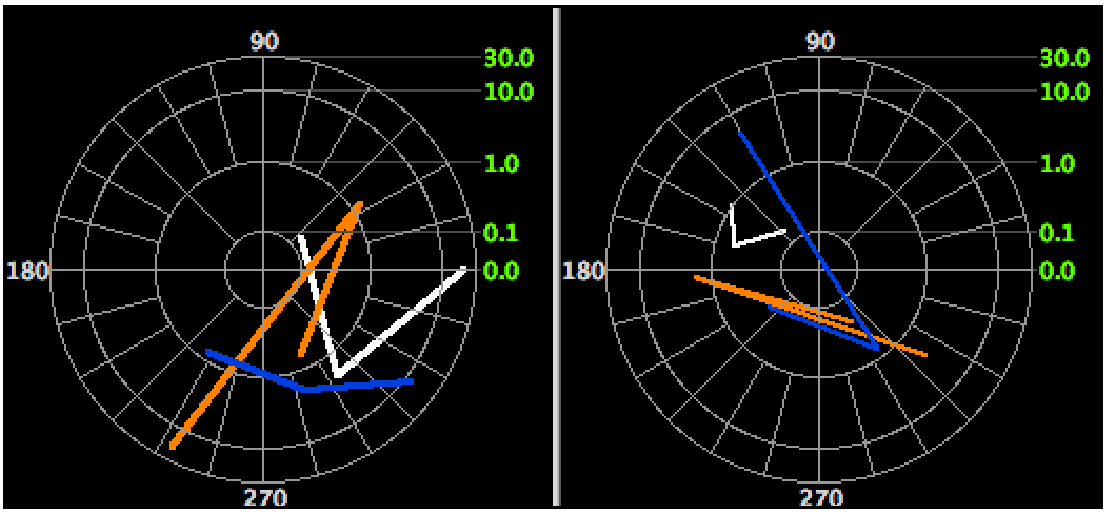


Figure 6.11. A first polar-log graph in *Lissajous-Fourier BCOM Trajectory* *.vi (level walking at 1.39 m/s, in a male aged 25 to 35).

For more details concerning polar logarithm graphs, see also chapter 9.

2.2. Discarded tests

In addition to the exceptions clearly illustrated in chapter 5 (par. 1.2.1 and 1.2.2), during the kinematic data analysis it became evident that other tests had to be rejected due to various and unexpected reasons. Consequently, some tests were discarded, particularly:

- level walking at 0.83 m/s was eliminated for 1 male and 1 female aged 6 to 13, 1 female aged 14 to 17, 1 male aged 25 to 35 and 1 female aged 56 to 65;
- level walking at 1.11 m/s was eliminated for 1 male aged 25 to 35 and 1 female aged 56 to 65;
- level walking at 1.39 m/s was eliminated for 1 male aged 18 to 24;
- level walking at 1.67 m/s was eliminated for 1 female aged 36 to 45;

- level walking at 1.94 m/s was eliminated for 1 female aged 25 to 35;
- level running at 1.94 m/s was eliminated for 1 female aged 18 to 24;
- level running at 2.22 m/s was eliminated for 1 female aged 18 to 24;
- level running at 2.50 m/s was eliminated for 1 female aged 6 to 13 and 1 female aged 18 to 24;
- level running at 2.78 m/s was eliminated for 2 females aged 18 to 24 and 1 female aged 25 to 35;
- level running at 3.06 m/s was eliminated for 1 male aged 56 to 65, 2 females aged 18 to 24 and 2 females aged 25 to 35.

On the whole, **22 trials** were deleted.

Furthermore, we had to discard tests for males and females aged 25 to 35, walking and running at different gradients:

- +5% walking at 0.83 m/s were eliminated for 2 males and 1 female;
- -5% walking at 0.83, 1.39, 1.67 and 1.94 m/s were eliminated for 1 female;
- -5% running at 1.94 m/s was eliminated for 1 male and 1 female;
- -10% walking at 0.83 m/s was eliminated for 2 females;
- -10% walking at 1.11 m/s was eliminated for 2 females;
- -10% running at 2.22 m/s was eliminated for 1 male;
- -10% running at 3.06 m/s was eliminated for 1 female;
- +15% walking at 0.83 m/s was eliminated for 1 female;
- +15% walking at 1.94 m/s was eliminated for 1 male;
- -15% walking at 1.11 m/s was eliminated for 1 female;
- +15% running at 3.06 m/s was eliminated for 1 female;
- -15% running at 2.78 m/s was eliminated for 1 male;
- +20% walking at 0.83 m/s was eliminated for 1 male and 2 females;
- +20% walking at 1.11 m/s was eliminated for 1 female;
- +20% walking at 1.39 m/s was eliminated for 1 male and 2 females;
- +20% walking at 1.67 m/s was eliminated for 2 females;
- +20% walking at 1.94 m/s was eliminated for 3 males;
- -20% walking at 1.39 m/s was eliminated for 1 female;
- -20% walking at 1.11, 1.67 and 1.94 m/s were eliminated for 1 male;
- +20% running at 2.50 and 2.78 m/s were eliminated for 1 female;
- -20% running at 2.22 and 2.78 m/s were eliminated for 1 male;

- +25% walking at 1.39 and 1.94 m/s were eliminated for 1 female;
- +25% walking at 1.67 m/s was eliminated for 2 females;
- +25% walking at 1.94 m/s was eliminated for 3 males;
- -25% walking at 1.11 and 1.94 m/s were eliminated for 1 male.

On the whole, **48 trials** were deleted.

Therefore, it is important to keep these information in mind when final results will be observed (see next chapters). In this case, average values will take into account these exceptions.

3. POSITION OF THE BODY CENTRE OF MASS AS A FUNCTION OF TIME

The file *.bcm format contains the periodic three-dimensional trajectory of the BCOM (see also chapter 1, par. 3.6.2). However, it is important to remember that the three-dimensional movement of the BCOM is always related to the variable time. Particularly, the movement time (both in walking and running) is the main variable that determines a different pattern in the BCOM displacement.

Therefore, before continuing, it is necessary to focus on the Fourier analysis. Indeed, this mathematical approach allowed us to find a significant relationship between the three-dimensional trajectory of the BCOM and the variable time.

Thus, the final step of kinematics analysis (from file *.bcm format to file *.res format) was based on this mathematical analysis.

4. FOURIER ANALYSIS

The mathematics of Fourier analysis has been used for many years to study physical phenomena coming from a wide variety of scientific and engineering fields (Schneider et al., 1983; Weaver, 1989; Crowe et al., 1996; Webb et al., 2007; Racic et al., 2009). Particularly, a continuous signal (meaning that the signal is present at all instances of time or space) is a time- or space- varying quantity that conveys information (Robertson et al., 2004). For instance, a simple method of Fourier analysis was used to analyze the forces exerted on the ground by men walking and running, by means of a force platform (Alexander, 1978). In this way, it could be possible to appreciate differences between walking and running, between slow and fast walking, between accelerated and decelerated walking, and finally between different individuals walking at the same speeds. Furthermore, derivatives and Fourier coefficients of human motion data were calculated, as well (Soudan et al., 1979).

Finally, the main benefits of Fourier analysis have been verified (Webb et al., 2007): a) to describe cyclical phenomena, including stride cycles; b) to describe the inter-limb coordination, since it describes each joint's contribution to limb movement; c) to compare strides of different duration; and d) to produce graphs of average joint movements for many strides and individuals.

In the following sections, we will briefly present the Fourier Series as a function. They can be placed in a complex exponential form that often simplifies the mathematical manipulations (Lamberto et al., 2000).

5. FOURIER SERIES

5.1. Fourier Series definition

Fourier Series are named in honor of Joseph Fourier (1768-1830), who made important contributions to the study of trigonometric series (Weaver, 1989; Enoka, 1994; O'Connor et al., 1996). Particularly, Fourier Series is the decomposition of a function in terms of sinusoidal functions (called *basis functions*) of different frequencies, that can be recombined to obtain the original function (Batschelet, 1981; Brockwell et al., 1991; Stokes, 1995; Alonso et al., 2005; Winter, 2005; Bruijn et al., 2009; Minetti, 2009). The recombination process is called *Fourier synthesis*. The result of this decomposition is the amount (e.g. amplitude) and the phase to be imparted to each basis function (or frequency) in the reconstruction. Therefore, it is also a function (of frequency), whose value could be represented as a complex number, in either polar or rectangular coordinates. Furthermore, it is referred to as the *frequency domain representation* of the original function (Batschelet, 1975; 1981; Dittrich et al., 2009; see par. 5.2 below).

In other words, the technique of Fourier analysis, which is only valid for periodic (cyclic) functions, involves the derivation of a series of sine and cosine terms to represent the frequency content of a signal (Batschelet, 1975; Enoka, 2002; Pecoraro et al., 2007; Racic et al., 2009). However, a non-periodic signal can be numerically converted to a periodic signal, and subjected to a Fourier analysis.

5.2. The frequency domain representation

5.2.1. Rectangular form

Any periodic waveform can be constructed by superimposing a combination of waveforms that have the proper amplitudes, phases and harmonics (Sutherland et al., 1980; Weaver, 1989; Giakas et al., 1997; Enoka, 2002; Racic et al., 2009).

Particularly, to analyze a periodic function, it is necessary to express the frequency content in terms of the ***fundamental frequency*** and its multiples (O'Connor et al., 1996; Lamberto et al., 2000; Winter, 2005; Figure 6.12).

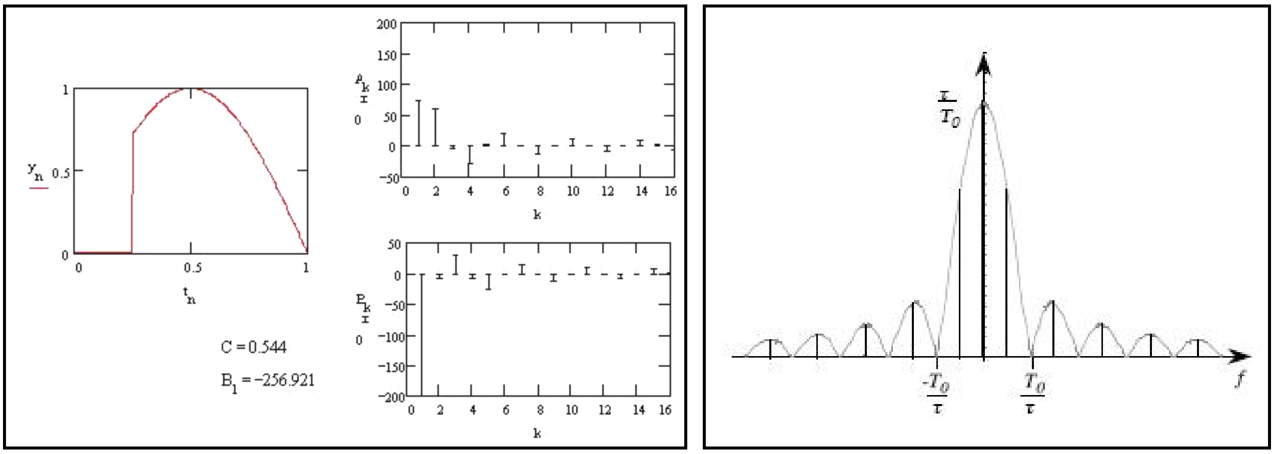


Figure 6.12. The relationship between a signal in the time domain (on the left) and its equivalent in the frequency domain (on the right).

Thus, the trigonometric Fourier Series representation of a periodic function is defined, in rectangular form, as:

$$y(t) = a_0 + \Sigma(a_i \bullet \cos\frac{2\pi i}{T} \bullet t + b_i \bullet \sin\frac{2\pi i}{T} \bullet t) \text{ [Eq. 6.4]}$$

where *y(t)* is the periodic function (or signal); a_0 the eventual constant term (Equation [6.5a]); a_i and b_i the cosine and sine Fourier Series coefficients (Equation [6.5b] and [6.5c], respectively; Batschelet, 1975; 1981; Weaver, 1989; Crowe et al., 1995; Diedrich et al., 1995; Doke et al., 2004; Winter, 2005; Racic et al., 2009); *T* the period; and ∑ the sum of cosine and sine functions.

$$a_0 = \int (y(t) \bullet \cos\frac{2\pi i}{T} \bullet t \bullet dt) \text{ [Eq. 6.5a]}$$

$$a_i = \frac{2}{T} \bullet \int (y(t) \bullet \cos\frac{2\pi i}{T} \bullet t \bullet dt) \text{ [Eq. 6.5b]}$$

$$b_i = \frac{2}{T} \bullet \int (y(t) \bullet \sin\frac{2\pi i}{T} \bullet t \bullet dt) \text{ [Eq. 6.5c]}$$

All these coefficients are defined in the range from -T/2 to T/2.

The absolute value of each coefficient reflects its importance in determining the over-all shape of the original waveform. Thus, the larger the value of the coefficient, the more effect it has in determining the shape of the waveform (Sutherland et al., 1980).

The fundamental frequency (Equation [6.6]) represents the single cosine + sine term, that best describes how the signal varies during one cycle (Antonsson et al., 1985; Enoka, 2002):

$$y(t) = a_1 \bullet \cos\frac{2\pi}{T} \bullet t + b_1 \bullet \sin\frac{2\pi}{T} \bullet t \quad [\text{Eq. } 6.6]$$

where a_1 and b_1 are the fundamental cosine and sine frequency coefficients, respectively.

The sine coefficients give the magnitudes of the waveforms that complete all the cycles during the movement and oscillate about the mean of the measurements, while the cosine coefficients give the magnitudes of waveforms that complete all the cycles during the movement, 90 degrees out of phase with the corresponding sinusoidal waveforms but otherwise identical in shape (Miller et al., 1973; Batschelet, 1975; 1981; McGinnis, 2005; Richards, 2008).

Multiples of the fundamental frequency are referred to as ***harmonics***:

$$y_n(t) = a_n \bullet \cos\frac{2\pi n}{T} \bullet t + b_n \bullet \sin\frac{2\pi n}{T} \bullet t \quad [\text{Eq. } 6.7]$$

where a_n and b_n are the n derivative cosine and sine frequency coefficients (Batschelet, 1981).

Once the fundamental frequency has been determined, the curve-fitting procedures are then used to determine the size of the respective harmonics that are needed to approximate the signal (Winter, 2005). In general, it is necessary to scale the contribution of each harmonic to the function. This contribution decreases as harmonic number increases for human movement; and it is weighted by means of a coefficient.

To sum up, a sinusoidal time-varying signal has four characteristics (Robertson et al., 2004):

1. offset or mean term (a_0), representing the average value of the signal (Batschelet, 1981);
2. amplitude (a or A), quantifying the magnitude of the oscillations;
3. frequency (f), representing how rapidly the signal oscillates;
4. phase angle (φ), the amount of time the signal may be delayed or time shifted.

5.2.2. Polar form

It is also possible to mathematically represent the Fourier Series of a periodic function in a polar form, as:

$$y(t) = C_0 + C_n \bullet \sin(t + \varphi_n) \quad [\text{Eq. } 6.8]$$

where C_0 represents the equivalent of a_0 (the eventual constant term); C_n and φ_n are respectively the coefficients harmonics (amplitudes) and the phases of the function, instead of a_i and b_i Fourier Series coefficients (Batschelet, 1975; 1981; Weaver, 1989; Hasan et al., 1996; Giakas et al., 1997; Racic et al., 2009).

This is the mathematical form (Equation [6.9]) we preferred and used to fully describe the three-dimensional displacement of the BCOM (Minetti, 2009):

$$x(t) = Ax1 \bullet \sin(t + \varphi 1) + Ax2 \bullet \sin(2t + \varphi 2) + Ax3 \bullet \sin(3t + \varphi 3) + Ax4 \bullet \sin(4t + \varphi 4) + Ax5 \bullet \sin(5t + \varphi 5) + Ax6 \bullet \sin(6t + \varphi 6)$$

[Eq. 6.9a]

$$y(t) = Ay1 \bullet \sin(t + \varphi 1) + Ay2 \bullet \sin(2t + \varphi 2) + Ay3 \bullet \sin(3t + \varphi 3) + Ay4 \bullet \sin(4t + \varphi 4) + Ay5 \bullet \sin(5t + \varphi 5) + Ay6 \bullet \sin(6t + \varphi 6)$$

[Eq. 6.9b]

$$z(t) = Az1 \bullet \sin(t + \varphi 1) + Az2 \bullet \sin(2t + \varphi 2) + Az3 \bullet \sin(3t + \varphi 3) + Az4 \bullet \sin(4t + \varphi 4) + Az5 \bullet \sin(5t + \varphi 5) + Az6 \bullet \sin(6t + \varphi 6)$$

[Eq. 6.9c]

where $x(t)$, $y(t)$ and $z(t)$ constitute the mathematical polar forms (or harmonics) of Fourier Series in forward, vertical and lateral direction, respectively; A represents the amplitude coefficient and φ the phase coefficient, in each movement direction.

To be precise, we decided to truncate the Fourier Series to the 6th harmonic according to Parseval's Theorem (Minetti, 2009; Parseval theorem in Mathematics - Wikipedia, the free encyclopedia, 2009) because:

- similar equations characterize well enough the real pattern of human locomotion;
- if this approach will be further used, the same Fourier coefficients and phases will be calculated. This is set out in Fourier analysis (Schneider et al., 1983; Weaver, 1989; Crowe et al., 1996; Pecoraro et al., 2007);
- coefficients past the sixth had very little influence of the final waveform (Sutherland et al., 1980). Moreover, it will be possible to add further harmonics without changing the values of the previous ones.

Furthermore, an important rule (see par. 2.1 above) that it is necessary to know in order to avoid disorders or errors in our data/results interpretation and discussion is that, in Fourier analysis:

- ***x-axis*** constitutes the anterior/posterior (or forward/backward) direction;
- ***y-axis*** the vertical direction;
- ***z-axis*** the medial/lateral direction.

5.2.3. Harmonic coefficients

Importantly, according to the double periodicity of the BCOM displacement (see chapter 1, par. 3.6.2), the **symmetrical coefficients** (Table 6.5, in which data refers to the example graphically reported in Figure 6.10 above) that characterize axes/directions in each harmonic are:

- in the forward and vertical directions: Ax2/φ2; Ax4/φ4; Ax6/φ6; Ay2/φ2; Ay4/φ4 and Ay6/φ6 (the so-called ‘odd’ coefficients);
- in the medial/lateral direction: Az1/φ1; Az3/φ3 and Az5/φ5.

MOVEMENT DIRECTION	SYMMETRICAL AMPLITUDES		SYMMETRICAL PHASES	
Forward	**Ax2**	12.0305	**φ2**	-0.6435
	Ax4	1.9112	**φ4**	-1.2325
	Ax6	0.7481	**φ6**	-2.1689
Vertical	**Ay2**	19.1163	**φ2**	-2.0459
	Ay4	1.3199	**φ4**	0.5984
	Ay6	0.6085	**φ6**	-1.1606
Medial/lateral	**Az1**	20.1055	**φ1**	0.0000
	Az3	2.0073	**φ3**	-0.9589
	Az5	0.1500	**φ5**	0.6727

Table 6.5. An example of symmetrical coefficients defining the individual Digital Locomotory Signature (level walking at 1.39 m/s, in a male aged 25 to 35).

However, according to the single periodicity of the BCOM displacement (see chapter 1, par. 3.6.2), the **asymmetrical coefficients** (Table 6.6, in which data refers to the example graphically reported in Figure 6.10 above) are:

- in the forward and vertical directions: Ax1/φ1; Ax3/φ3; Ax5/φ5; Ay1/φ1; Ay3/φ3 and Ay5/φ6 (the so-called ‘even’ coefficients);
- in the medial/lateral direction: Az2/φ2; Az4/φ4 and Az6/φ6.

MOVEMENT DIRECTION	ASYMMETRICAL AMPLITUDES		ASYMMETRICAL PHASES	
Forward	**Ax1**	4.6553	**φ1**	2.0885
	Ax3	0.7684	**φ3**	-0.9385
	Ax5	0.2175	**φ5**	-2.4914
Vertical	**Ay1**	2.5532	**φ1**	-0.6780
	Ay3	1.6231	**φ3**	-3.0747
	Ay5	0.2277	**φ5**	-1.0192
Medial/lateral	**Az2**	0.9686	**φ2**	2.5406
	Az4	0.5121	**φ4**	2.8675
	Az6	0.1576	**φ6**	2.3028

Table 6.6. An example of asymmetrical coefficients defining the individual Digital Locomotory Signature (level walking at 1.39 m/s, in a male aged 25 to 35).

Therefore, referring to data reported in Figure 6.10 (and Tables 6.5 and 6.6), specific **symmetrical harmonics** of such BCOM displacement are:

$$x = 12.0305 \bullet (2t - 0.6453) + 1.9112 \bullet (4t - 1.2325) + 0.7481 \bullet (6t - 2.1689) \quad \text{[Eq. 6.10a]}$$

$$y = 19.1163 \bullet (2t - 2.0459) + 1.3199 \bullet (4t + 0.5984) + 0.6085 \bullet (6t - 1.1606) \quad \text{[Eq. 6.10b]}$$

$$z = 20.1055 \bullet (t + 0.0000) + 2.0073 \bullet (3t - 0.9589) + 0.1500 \bullet (5t + 0.6727) \quad \text{[Eq. 6.10c]}$$

However, specific **asymmetrical harmonics** are:

$$x = 4.6553 \bullet (t + 2.0885) + 0.7684 \bullet (3t - 0.9385) + 0.2175 \bullet (5t - 2.4914) \quad \text{[Eq. 6.10d]}$$

$$y = 2.5532 \bullet (t - 0.6780) + 1.6231 \bullet (3t - 3.0747) + 0.2277 \bullet (5t - 1.0192) \quad \text{[Eq. 6.10e]}$$

$$z = 0.9686 \bullet (2t + 2.5046) + 0.5121 \bullet (4t + 2.8675) + 0.1576 \bullet (6t + 2.3028) \quad \text{[Eq. 6.10f]}$$

Importantly, the combination of these sets of equation together define and characterize the so-called *Digital Locomotory Signature* (DLS), a 3D pattern of BCOM dynamics. It will be widely described in chapter 7.

6. TRAJECTORY OF THE BODY CENTRE OF MASS OVER SPACE AND TIME

6.1. Introduction

Most of human motion is periodic (walking and running, in particular; Lestrel et al., 1977; Alexander et al., 1978; Soudan et al., 1979; Alexander et al., 1980; Sutherland et al., 1980; McDonald's et al., 1990; Lestrel et al., 2005). Transformations representing the spatial-temporal characteristics of periodic trajectories could be used to describe some functions involving motion.

The individual three-dimensional trajectory of the BCOM while moving on a treadmill (walking and running) is then a periodic function, mathematically described by harmonics characterizing the Digital Locomotory Signature (see par. 5.2.3 above).

It could be represented as a closed loop (Lissajous *contour*; Lissajous figures in Mathematics - Wikipedia, the free encyclopedia, 2009), following the same pattern at each stride.

In Mathematics, Lissajous *contours* (Emmerton, 1986) are the family of curves described by the parametric equations:

$$x(t) = A \bullet (\cos \omega_x t - \varphi_x) \quad \text{[Eq. 6.11a]}$$

$$y(t) = B \bullet (\cos \omega_y t - \varphi_y) \quad \text{[Eq. 6.11b]}$$

which describe complex harmonic motion; sometimes also written in the forms:

$$x(t) = A \bullet (\sin \omega t + \varphi) \quad \text{[Eq. 6.12a]}$$

$$y(t) = B \bullet (\sin t) \quad \text{[Eq. 6.12b]}$$

where *A* and *φ* are the coefficient amplitude and the phase, respectively (see par. 5.2.2 above).

The appearance of the curve is highly sensitive to the ratio a/b. For a ratio of 1, the figure is an ellipse, with special cases including circles ($A = B$; $\varphi = \pi/2$ radians) and lines ($\varphi = 0$). Another simple Lissajous figure is the parabola ($a/b = 2$; $\varphi = \pi/2$). Other ratios produce more complicated curves, which are closed only if a/b is rational (Figure 6.13a, and 6.13b), where, respectively, a is 1, 5 and 9; and b is 2, 6 and 8.

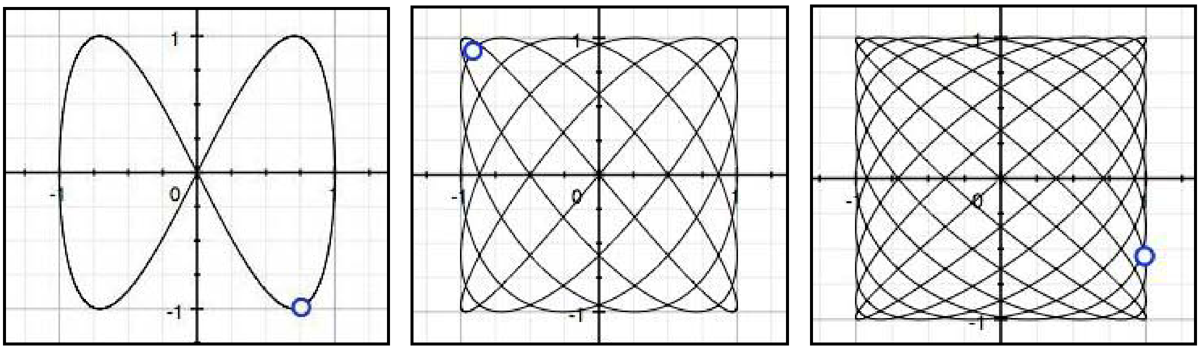


Figure 6.13a. Some simple Lissajous curves in 2D.

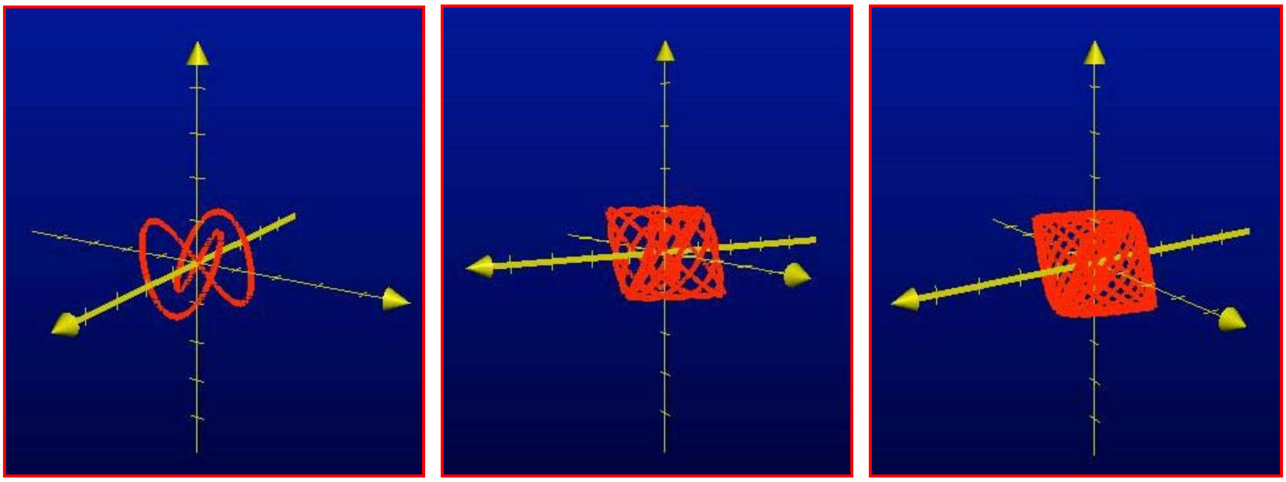


Figure 6.13b. Some simple Lissajous curves in 3D.

6.2. Lissajous *contours* in our study

As it has been already presented, Lissajous *contours* (in 3D) are very important in the graphical representation of periodic functions (i.e. three-dimensional displacement of the BCOM). In fact, they summarize the characteristics of Fourier Series, using them to graphically describe these continuous functions (Minetti, 2009). Therefore, it is important to focus on this significant coupling.

Indeed, this approach will provide a different view angle to the mechanical paradigm of locomotion. To be precise, as shown, the individual 3D trajectory of BCOM while moving on a treadmill is a closed loop (Lissajous *contour*) travelled by it at each stride. This is the consequence of all its raisings, forward-backward and lateral displacements (Minetti, 2006; 2009). Furthermore, in the obtained closed loops, it is possible to clearly represent when in the stride there is single and double contact, in addition to the eventual flight phase. Finally, the mathematical method (based on Fourier Series and Lissajous analysis) could allow us to fully describe the individual behaviour of the BCOM.

Consequently, all *contours* presented and discussed in the chapter 7 have been drawn up by following the mathematical and graphical approach described in this chapter.

7. CONCLUSION

In this chapter, the single steps to analyse kinematic data have been deeper presented and discussed. Importantly, we develop both a method and a valid evaluation protocol to explain (mathematically and graphically) the 3D displacement of the BCOM. As a result, focusing on the BCOM as the investigation object fulfilling such a need, this has been achieved through a different use of classic biomechanical procedures.

Thus, in the following chapters, we will investigate in all testing conditions:

- the pattern of both amplitudes and phases in characterizing the Digital Locomotory Signature, by investigating average 3D *contours* as gender, age, gait, speed and gradient change (chapter 7). Moreover, we will briefly focus on the excursion of the BCOM (Appendix 7.2);
- the role of (symmetrical and asymmetrical) amplitude coefficients in defining the Symmetry Index, namely representing the spatial differences, in BCOM trajectory, between two steps (chapter 8);
- the graphical representation of both amplitudes and phases in a polar graph (chapter 9).

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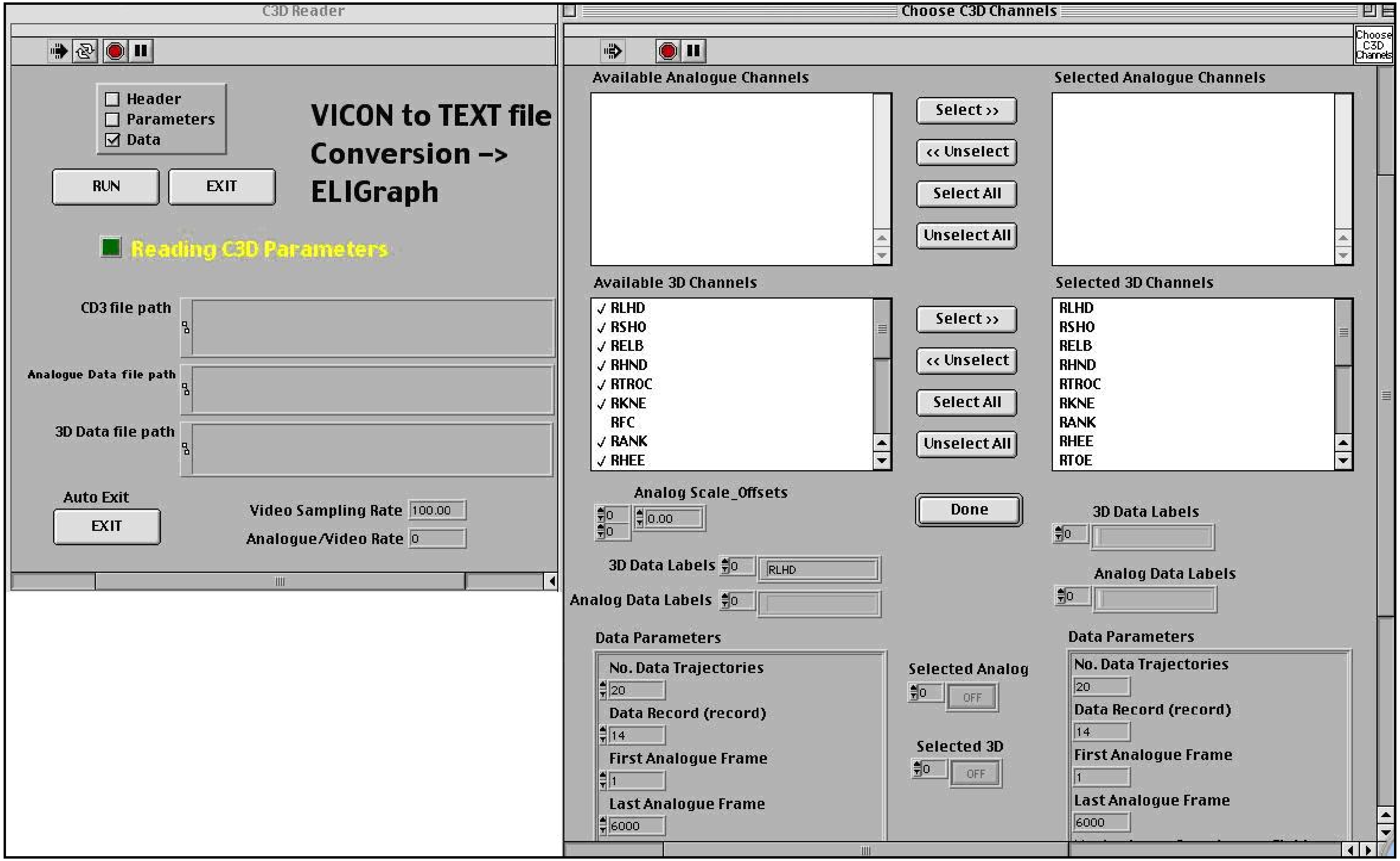


☐ Header
☐ Parameters
☑ Data

VICON to TEXT file Conversion -> ELIGraph

🟩 Reading C3D Parameters

CD3 file path
Analogue Data file path
3D Data file path

Video Sampling Rate 100.00
Analogue/Video Rate 0

Available Analogue Channels

Selected Analogue Channels

Available 3D Channels
✓ RLHD
✓ RSHO
✓ RELB
✓ RHND
✓ RTROC
✓ RKNE
RFC
✓ RANK
✓ RHEE

Selected 3D Channels
RLHD
RSHO
RELB
RHND
RTROC
RKNE
RANK
RHEE
RTOE

Analog Scale_Offsets 0 0 0.00

3D Data Labels 0 RLHD
Analog Data Labels 0

Data Parameters
No. Data Trajectories 20
Data Record (record) 14
First Analogue Frame 1
Last Analogue Frame 6000

Selected Analog 0 OFF
Selected 3D 0 OFF

3D Data Labels 0
Analog Data Labels 0

Data Parameters
No. Data Trajectories 20
Data Record (record) 14
First Analogue Frame 1
Last Analogue Frame 6000

Figure 6.1. *C3D Reader* *.vi, in LabVIEW 6.1.

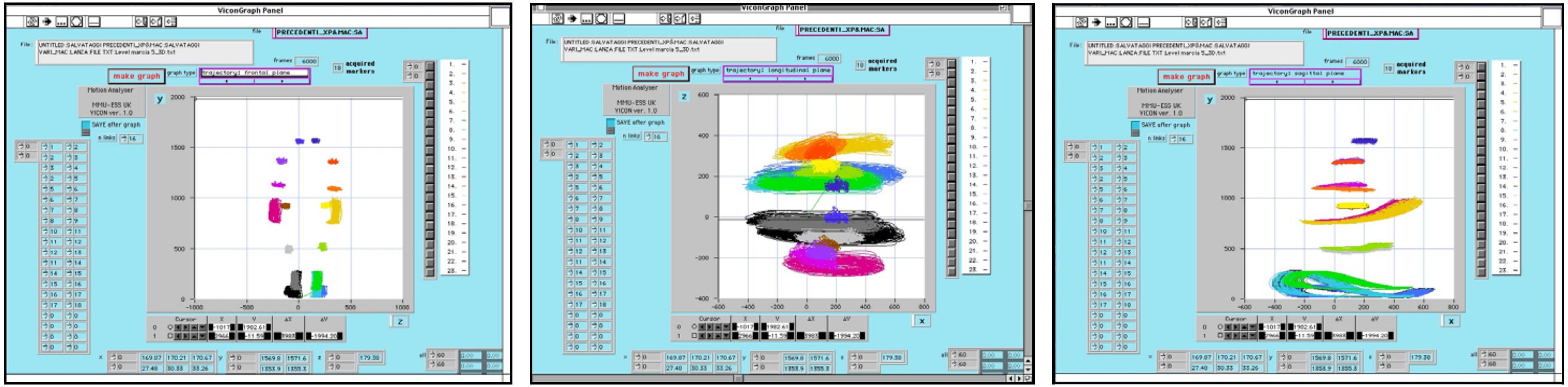


Figure 6.2. Movement of the 18 markers in the frontal (on the left), longitudinal (in the middle) and sagittal (on the right) plane.

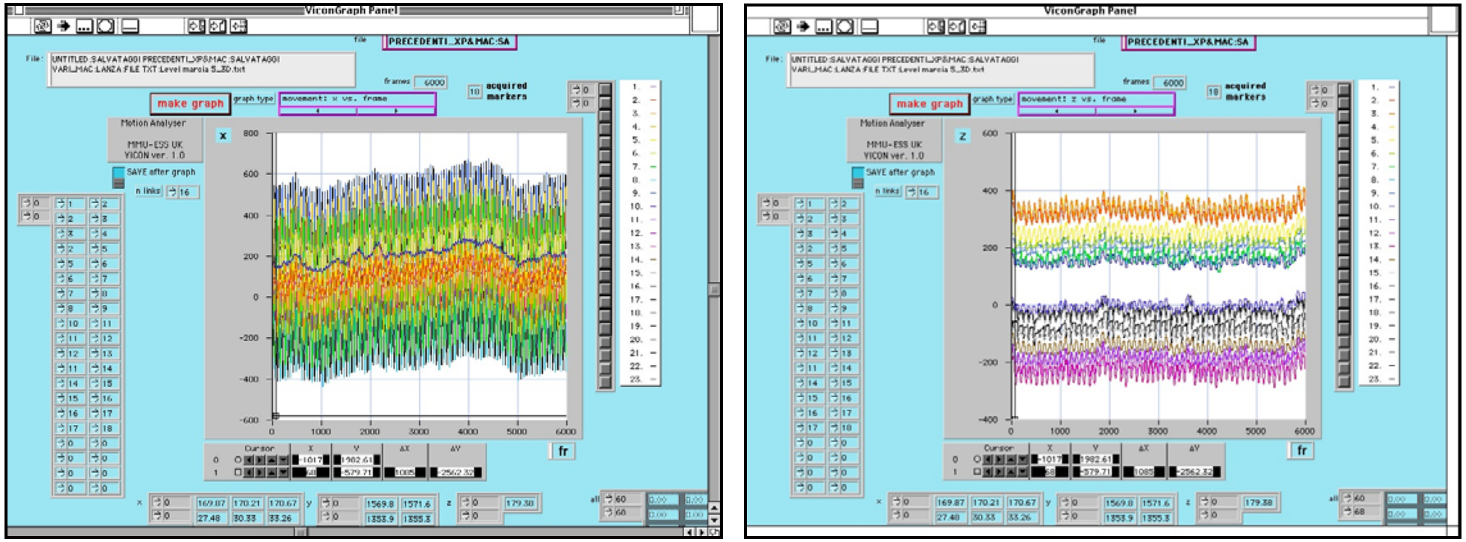


Figure 6.3. Anterior/posterior (on the left) and medial/lateral (on the right) movements, visualised by *ViconGraph* *.vi.

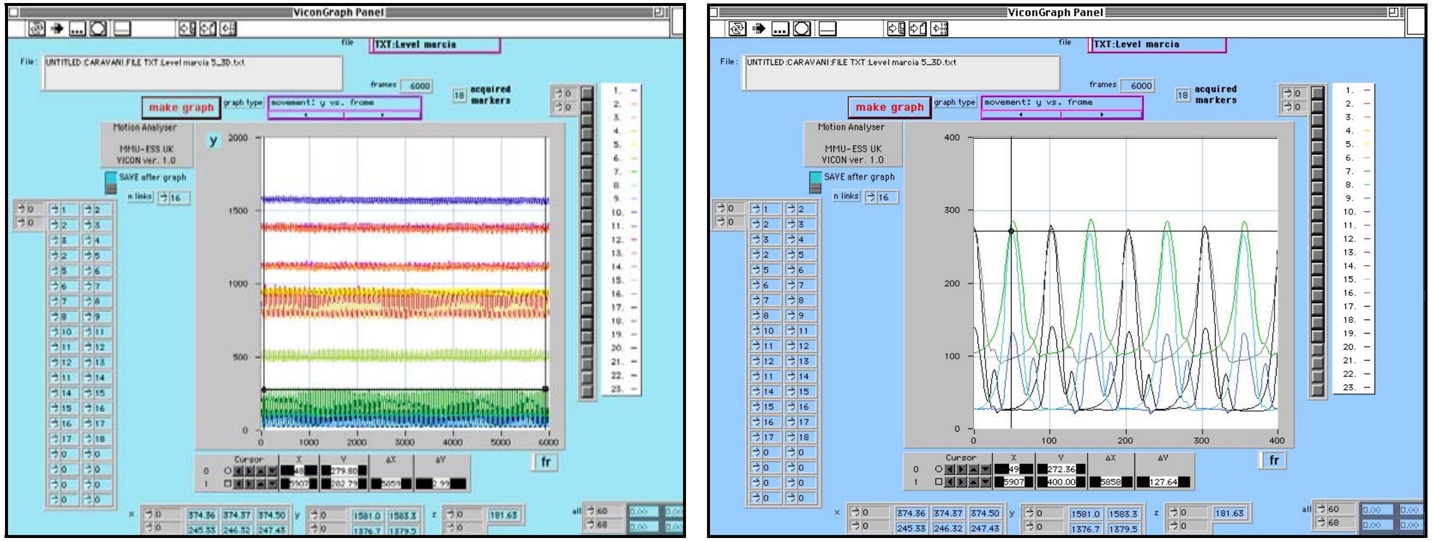
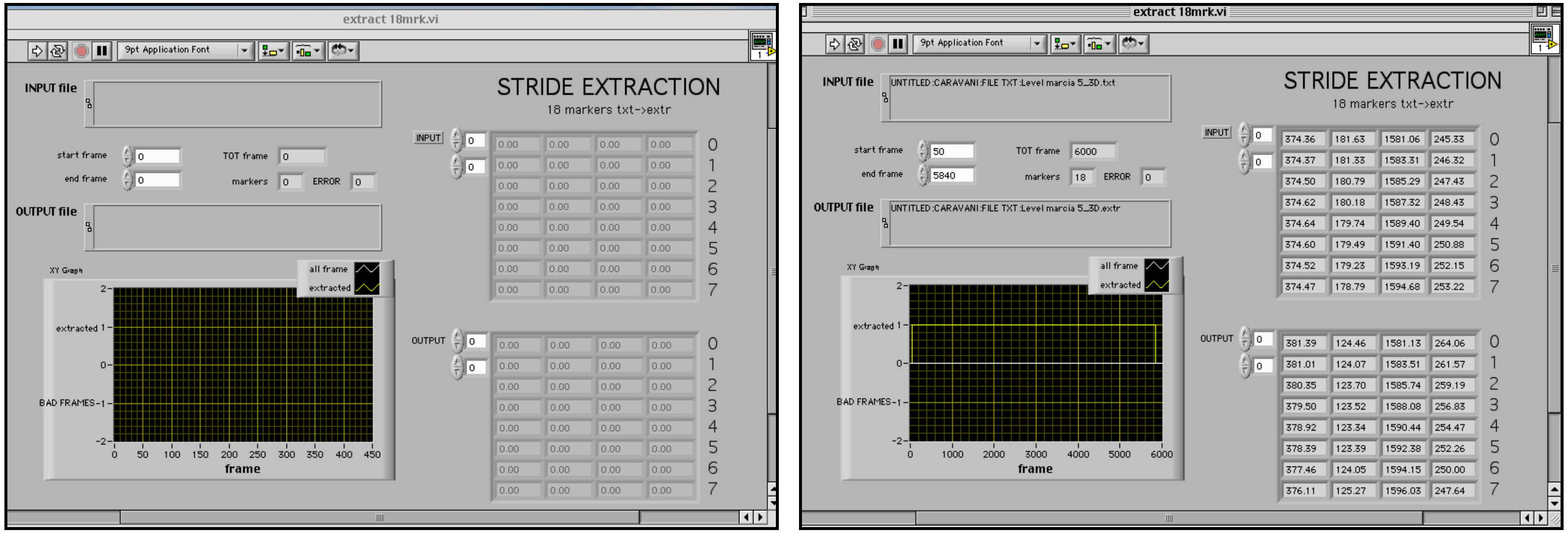


Figure 6.4. Vertical movement, visualised by *ViconGraph* *.vi.



STRIDE EXTRACTION

18 markers txt->extr

INPUT file: UNTITLED:CARAVANI:FILE TXT:Level marcia 5_3D.txt

start frame 50

end frame 5840

TOT frame 6000

markers 18

ERROR 0

OUTPUT file: UNTITLED:CARAVANI:FILE TXT:Level marcia 5_3D.extr

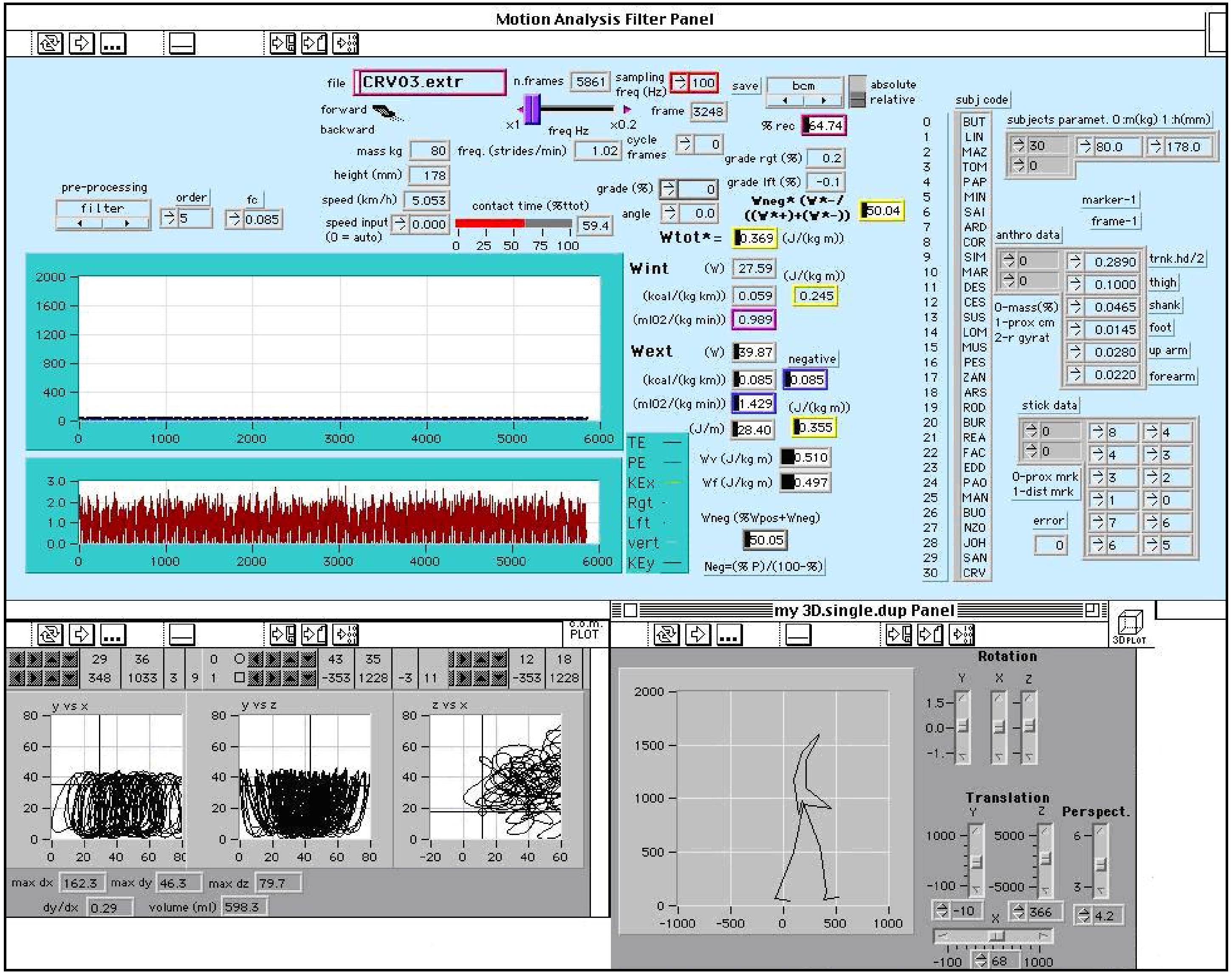
INPUT 0 0

374.36	181.63	1581.06	245.33	0
374.37	181.33	1583.31	246.32	1
374.50	180.79	1585.29	247.43	2
374.62	180.18	1587.32	248.43	3
374.64	179.74	1589.40	249.54	4
374.60	179.49	1591.40	250.88	5
374.52	179.23	1593.19	252.15	6
374.47	178.79	1594.68	253.22	7

OUTPUT 0 0

381.39	124.46	1581.13	264.06	0
381.01	124.07	1583.51	261.57	1
380.35	123.70	1585.74	259.19	2
379.50	123.52	1588.08	256.83	3
378.92	123.34	1590.44	254.47	4
378.39	123.39	1592.38	252.26	5
377.46	124.05	1594.15	250.00	6
376.11	125.27	1596.03	247.64	7

Figure 6.5. *Extract 18 mrk* *.vi, in LabVIEW 6.1.



file CRV03.extr

n.frames 5861

sampling freq (Hz) 100

save bcm

frame 3248

% rec 64.74

mass kg 80

freq. (strides/min) 1.02

cycle frames 0

height (mm) 178

speed (km/h) 5.053

speed input (0 = auto) 0.000

contact time (%ttot) 59.4

grade (%) 0

angle 0.0

grade rgt (%) 0.2

grade lft (%) -0.1

Wneg* (W*-/((W*+)+(W*-)) 50.04

Wtot*= 0.369 (J/(kg m))

pre-processing filter

order 5

fc 0.085

Wint (W) 27.59

(kcal/(kg km)) 0.059

(J/(kg m)) 0.245

(ml02/(kg min)) 0.989

Wext (W) 39.87

(kcal/(kg km)) 0.085

negative 0.085

(ml02/(kg min)) 1.429

(J/(kg m)) 0.355

(J/m) 28.40

Wv (J/kg m) 0.510

Wf (J/kg m) 0.497

Wneg (%Wpos+Wneg) 50.05

Neg=(% P)/(100-%)

subj code

0	BUT
1	LIN
2	MAZ
3	TOM
4	PAP
5	MIN
6	SAI
7	ARD
8	COR
9	SIM
10	MAR
11	DES
12	CES
13	SUS
14	LOM
15	MUS
16	PES
17	ZAN
18	ARS
19	ROD
20	BUR
21	REA
22	FAC
23	EDD
24	PAO
25	MAN
26	BUO
27	NZO
28	JOH
29	SAN
30	CRV

subjects paramet. 0:m(kg) 1:h(mm) 30 80.0 178.0 0

marker-1

frame-1

anthro data 0 0

0-mass(%) 1-prox cm 2-r gyrat

0.2890	trnk.hd/2
0.1000	thigh
0.0465	shank
0.0145	foot
0.0280	up arm
0.0220	forearm

stick data 0 0

0-prox mrk 1-dist mrk

8	4
4	3
3	2
1	0
7	6
6	5

error 0

29 36 0 43 35 12 18

348 1033 3 9 1 -353 1228 -3 11 -353 1228

max dx 162.3 max dy 46.3 max dz 79.7

dy/dx 0.29 volume (ml) 598.3

-10 366 4.2 68

Figure 6.6. *Motion Analysis Filter* *.vi, in LabVIEW 2.2.1.

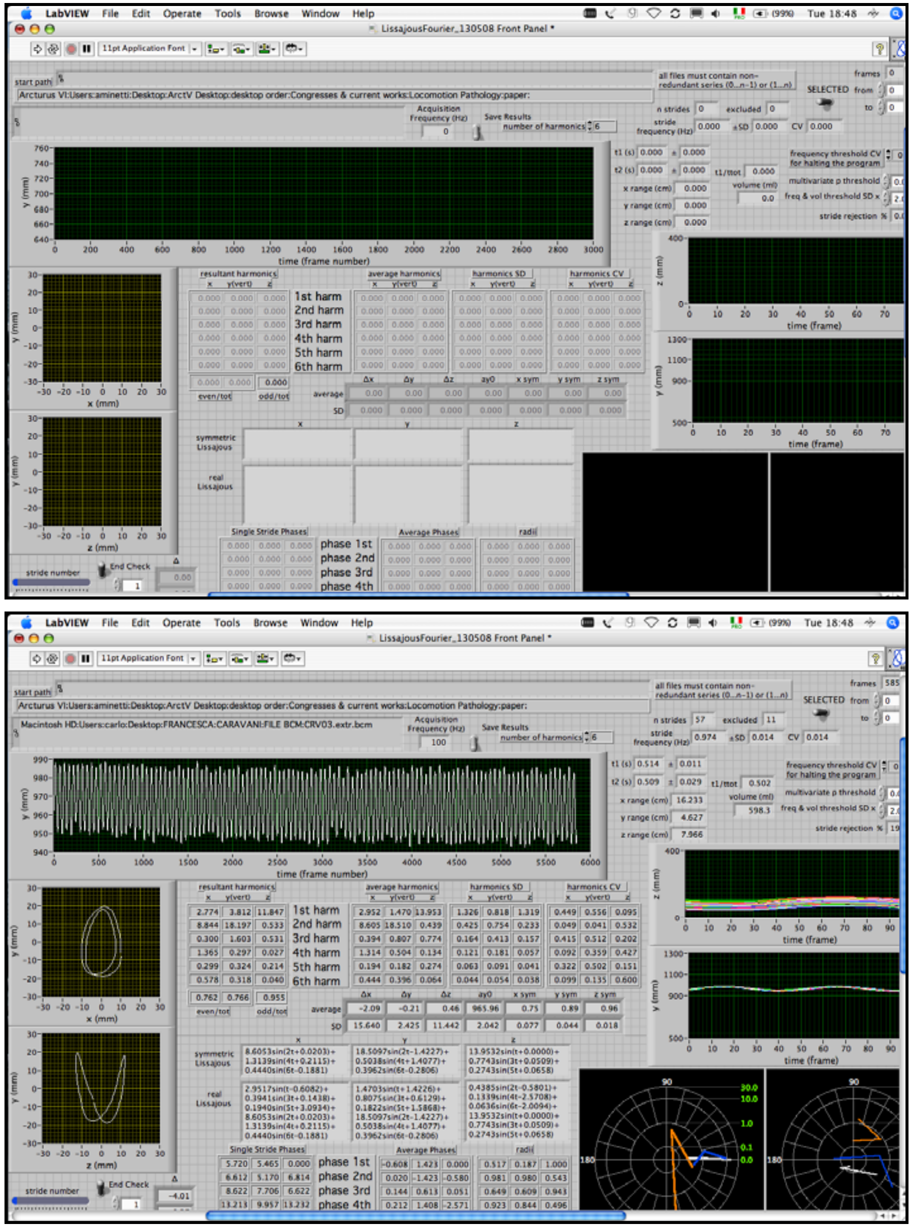


Figure 6.8. *Lissajous-Fourier BCOM Trajectory* *.vi, in LabVIEW 7.1.

PART 2

RESULTS AND DISCUSSION

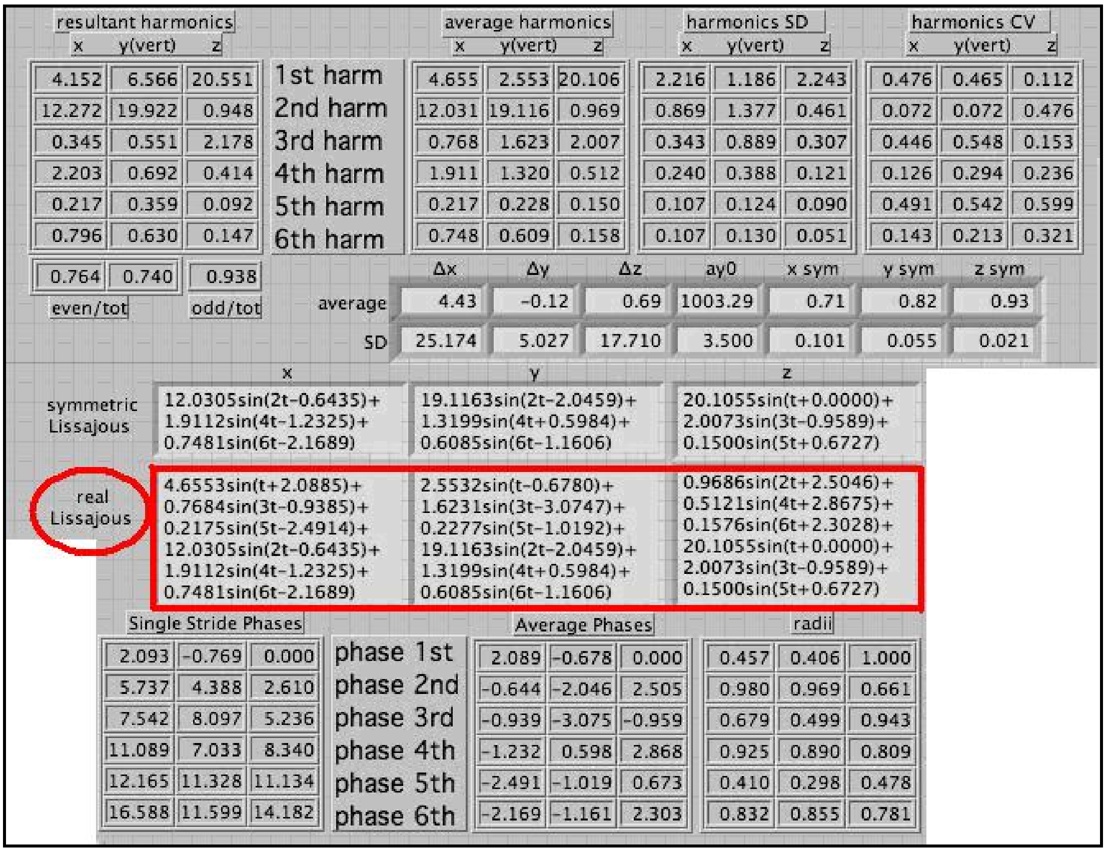
Chapter 7

DIGITAL LOCOMOTORY SIGNATURE

1. FOURIER ANALYSIS AND LISSAJOUS *CONTOURS*

As already widely described in chapter 6, each 3D trajectory of the BCOM specific of gender, age, type of locomotion, speed and gradient could be mathematical represented by means of a method using Fourier analysis and graphically represented by Lissajous *contours* (Minetti, 2009).

As a result, discrete kinematic variables were then converted into corresponding functions (continuous, by definition). In particular, in each trial the displacement of the BCOM (Figure 7.1) is characterized by: a) 1 amplitude coefficient (A) and 1 phase (φ) *per* each axis/rectangular component; and b) 6 harmonics *per* each of the 3 spatial coordinates of the BCOM (forward, vertical and lateral), with time as the independent variable.



resultant harmonics				average harmonics			harmonics SD			harmonics CV		
x	y(vert)	z		x	y(vert)	z	x	y(vert)	z	x	y(vert)	z
4.152	6.566	20.551	1st harm	4.655	2.553	20.106	2.216	1.186	2.243	0.476	0.465	0.112
12.272	19.922	0.948	2nd harm	12.031	19.116	0.969	0.869	1.377	0.461	0.072	0.072	0.476
0.345	0.551	2.178	3rd harm	0.768	1.623	2.007	0.343	0.889	0.307	0.446	0.548	0.153
2.203	0.692	0.414	4th harm	1.911	1.320	0.512	0.240	0.388	0.121	0.126	0.294	0.236
0.217	0.359	0.092	5th harm	0.217	0.228	0.150	0.107	0.124	0.090	0.491	0.542	0.599
0.796	0.630	0.147	6th harm	0.748	0.609	0.158	0.107	0.130	0.051	0.143	0.213	0.321

even/tot		odd/tot
0.764	0.740	0.938

	Δx	Δy	Δz	ay0	x sym	y sym	z sym
average	4.43	-0.12	0.69	1003.29	0.71	0.82	0.93
SD	25.174	5.027	17.710	3.500	0.101	0.055	0.021

	x	y	z
symmetric Lissajous	12.0305sin(2t-0.6435)+ 1.9112sin(4t-1.2325)+ 0.7481sin(6t-2.1689)	19.1163sin(2t-2.0459)+ 1.3199sin(4t+0.5984)+ 0.6085sin(6t-1.1606)	20.1055sin(t+0.0000)+ 2.0073sin(3t-0.9589)+ 0.1500sin(5t+0.6727)
real Lissajous	4.6553sin(t+2.0885)+ 0.7684sin(3t-0.9385)+ 0.2175sin(5t-2.4914)+ 12.0305sin(2t-0.6435)+ 1.9112sin(4t-1.2325)+ 0.7481sin(6t-2.1689)	2.5532sin(t-0.6780)+ 1.6231sin(3t-3.0747)+ 0.2277sin(5t-1.0192)+ 19.1163sin(2t-2.0459)+ 1.3199sin(4t+0.5984)+ 0.6085sin(6t-1.1606)	0.9686sin(2t+2.5046)+ 0.5121sin(4t+2.8675)+ 0.1576sin(6t+2.3028)+ 20.1055sin(t+0.0000)+ 2.0073sin(3t-0.9589)+ 0.1500sin(5t+0.6727)

Single Stride Phases				Average Phases			radii		
2.093	-0.769	0.000	phase 1st	2.089	-0.678	0.000	0.457	0.406	1.000
5.737	4.388	2.610	phase 2nd	-0.644	-2.046	2.505	0.980	0.969	0.661
7.542	8.097	5.236	phase 3rd	-0.939	-3.075	-0.959	0.679	0.499	0.943
11.089	7.033	8.340	phase 4th	-1.232	0.598	2.868	0.925	0.890	0.809
12.165	11.328	11.134	phase 5th	-2.491	-1.019	0.673	0.410	0.298	0.478
16.588	11.599	14.182	phase 6th	-2.169	-1.161	2.303	0.832	0.855	0.781

Figure 7.1. In the red box, harmonic amplitudes and phases (= 36 values defining the so-called *Digital Locomotory Signature*), in level walking at 1.39 m/s, in a male aged 25 to 35.

To sum up, each 3D trajectory of the BCOM has been described by 36 values together defining the so-called *Digital Locomotory Signature* (DLS).

$$DLS = 36VALUES = 2coefficients \bullet 6harmonics \bullet 3spatialcoordinates \quad [Eq.\ 7.1]$$

Single values of both symmetrical and asymmetrical coefficients (males and females in all age group, in both gaits) are contained in the enclosed CD (First Study, Chapter 7, Template level gait and Template gradient gait).

2. DIGITAL LOCOMOTORY SIGNATURE

2.1. Introduction

The so-called *Digital Locomotory Signature* has been mathematically defined for each gender, age group, type of locomotion, speed and gradient. It has been graphically represented by *Grapher* (APPLE, USA), on a Macintosh Notebook.

According to the right-hand rule and to what previously stated in chapter 6 (par. 2.1), it is important to note that, in *Grapher*: a) ***x-axis*** constitutes the anterior/posterior (or forward/backward) direction; b) ***y-axis*** the medial/lateral direction. Importantly, right and left sides are always inverted; and c) ***z-axis*** the vertical direction.

Firstly, in each testing condition, DLS has been qualitatively analysed (par. 2.2).

Secondly, based upon this preliminary approach, a deeper statistical analysis on the main harmonic coefficients has been performed (par. 2.3).

Average 3D *contours* are contained in the enclosed CD (First Study, Chapter 7, Grapher 3D *contours*: as a function of age; as a function of speed; and as a function of gradient).

2.2. Qualitative analysis

Average values of continuous functions were calculated *per* weighing up the effect of each variable: gender, age, speed and gradient.

Finally, these average *contours* were evaluated using a qualitative approach, and they were compared in the different testing conditions.

This preliminary analysis underlines that, at the level gradient, both in walking and running, the most significant differences occur similarly in males and females between:

1. children aged 6 to 13 (Lefebvre et al., 2002);
2. young adults aged 25 to 35;
3. elderly adults aged 56 to 65.

The pattern of DLS in the other age groups (from 14 to 55 years) seems not to significantly differ from young adults.

Therefore, we mainly investigated the pattern of DLS in these three age groups.

2.3. Statistical analysis

In order to complete the simplest qualitative analysis, a more complex (and objective) statistical analysis has been performed, as well. As stated before, only the three main age groups were considered (6 to 13; 25 to 35; and 56 to 65 years). Specifically, we wanted to find out whether significant differences exist in harmonic coefficients, both in males and females, as age, type of locomotion and speed vary.

Therefore, a one-way ANOVA with Huynh-Feldt correction for repeated measures were used to assess, in each movement direction, the main amplitude differences, among the three age groups, between gender (males *versus* females) and speed, and the possible interaction among these two variables. Finally, it is important to note that the previously described statistical analysis could not be satisfactorily applied to phase coefficients. Indeed, they constitute a circular variable differently to amplitudes (see also chapter 9, par. 2.2 and 2.3). In this case, a circular statistical analysis has to be performed. However, we have decided to consider the corresponding sine and cosine functions (ranging from -1 to 1) to solve this problem (see also chapter 6). Indeed, both sine and cosine functions are linear variables on which a one-way ANOVA with Huynh-Feldt correction for repeated measures could be applied in order to assess the main differences, between gender and speed, and the possible interaction among these two variables.

In detail, the last two coefficients ($A5/\varphi5$ and $A6/\varphi6$) were not considered in these statistical analyses because of their relative importance in the characterization of the Digital Locomotory Signature (according to Parseval's Theorem: see chapter 6, par. 5.2.2). However, they were used in the graphical representation of the closed loops (qualitative analysis).

Results of these statistical analyses have been widely illustrated and discussed in Appendix 7.1, at the end of the chapter. Moreover, as previously shown in par. 1, single amplitudes and phases are contained in the enclosed CD (First Study, Chapter 7, Coefficients statistical analysis). In addition, in the section Coefficients Regression Analysis, results (in terms of Coefficient of Determination and Coefficient of Correlation) have been presented as a function of gender, age, gaits, speed and gradient. To be more precise, statistical significance has been underlined in different colours.

2.4. Graph legend

Graphs (by using *Grapher*) were drawn up as follows:

- values in level/gradient walking are plotted in blue at the speed of 0.83 m/s; in yellow at the speed of 1.11 m/s; in dark green at the speed of 1.39 m/s; in ski-blue at the speed of 1.67 m/s; and in red at the speed of 1.94 m/s;

- values in level/gradient running are plotted in green at the speed of 1.94 m/s; in orange at the speed of 2.22 m/s; in white at the speed of 2.50 m/s; in pink at the speed of 2.78 m/s; and in blue at the speed of 3.06 m/s;
- furthermore, values are plotted in white for males and females aged 6 to 13; in red for subjects aged 14 to 17; in dark green for subjects aged 18 to 24; in orange for subjects aged 25 to 35; in ski-blue for subjects aged 36 to 45; in blue for subjects aged 46 to 55; and in pink for subjects aged 56 to 65.

3. DIGITAL LOCOMOTORY SIGNATURE AS A FUNCTION OF GENDER

3.1. Introduction

In the following sections, there are 3D *contours*, at the level gradient, in both walking and running (at all different speeds), in males (on the left) and females (on the right) of the most representative age groups: subjects aged 6 to 13, 25 to 35, and 56 to 65.

To be precise, a graphical rear view of the average BCOM pattern has been proposed in each testing condition; moreover, the graphical limits in each direction have been defined.

3.2. Results of our experiments

3.2.1. Level walking at all speeds

A. Both in males and females **aged 6 to 13** (Figure 7.2), limits of each graph are: a) -30/30 mm along the forward direction; b) 710/810 mm along the vertical direction (according to Ay0 values; see also chapter 6, par. 2.1); and c) -30/30 mm along the lateral direction.

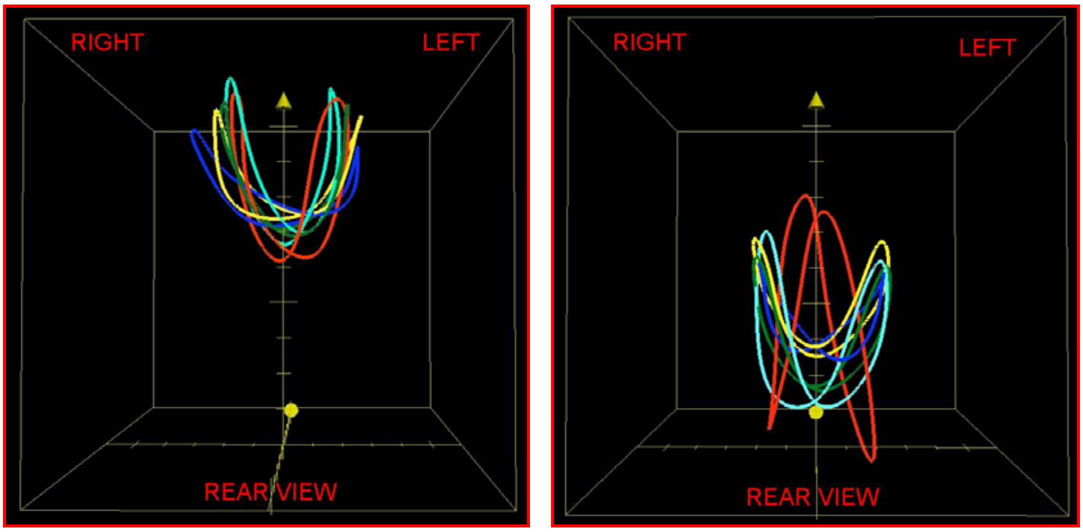


Figure 7.2. 3D *contours* as a function of all walking speeds, males (on the left) and females aged 6 to 13 (on the right).

As shown in these *contours*, on average, the BCOM is slightly raised and lifted (i.e. a higher vertical position) in males because their higher anthropometric dimensions (i.e. height) compared to females (see also chapter 5, par. 1.3; Song et al., 1997). In both males and females, the BCOM becomes more vertical with increasing walking speed (Beauchet et al., 2009a; 2009b). However, in females, the highest speeds (1.67 and 1.94 m/s) seem to have the most strange pattern: this is probably related to the subjects difficulty to maintain this speed for a long time (1' minute).

B. Both in males and females **aged 25 to 35** (Figure 7.3), limits of each graph are: a) -30/30 mm along the forward direction; b) 910/1030 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

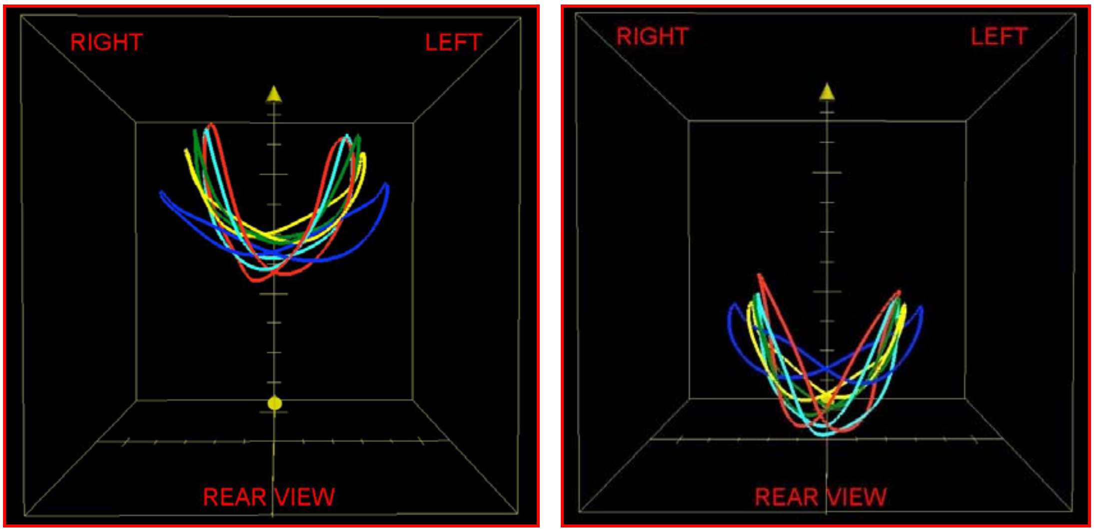


Figure 7.3. 3D *contours* as a function of all walking speeds, males (on the left) and females aged 25 to 35 (on the right).

On average, the BCOM is slightly raised and lifted in males because their higher heights compared to females. Evidently, in both males and females, the BCOM becomes more vertical with increasing walking speed. However, in females, it seems that this vertical organization is faster than in males: this is probably related to their different pelvis configuration (see also Appendix 7.2).

C. Both in males and females **aged 56 to 65** (Figure 7.4), limits of each graph are: a) -30/30 mm along the forward direction; b) 870/1030 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

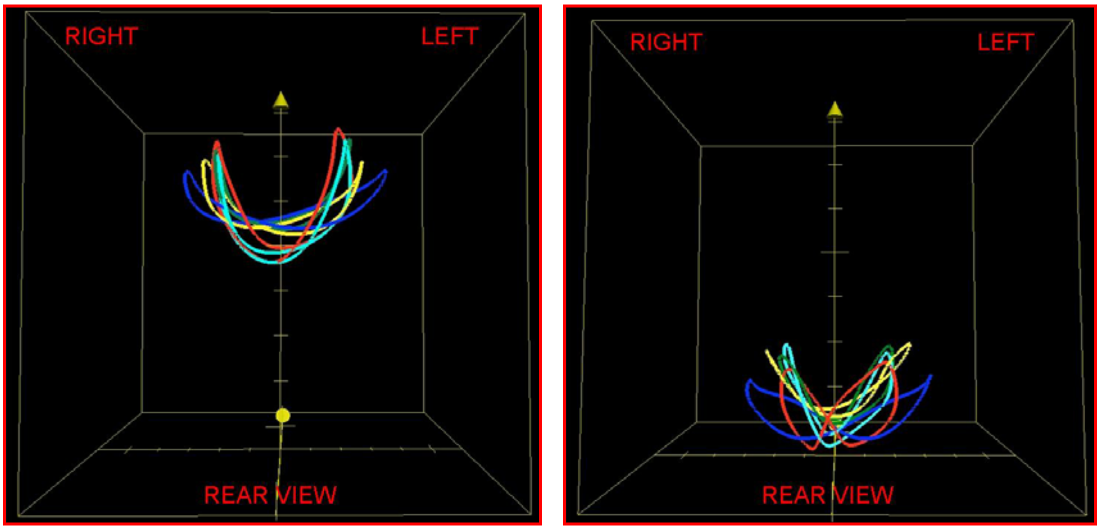


Figure 7.4. 3D *contours* as a function of all walking speeds,
males (on the left) and females aged 56 to 65 (on the right).

On average, the BCOM is slightly raised and lifted in males because their higher heights compared to females. Evidently, in both males and females, the BCOM becomes slightly more vertical with increasing walking speed. However, in females, it seems that this vertical organization is faster than in males: this is probably related to their different pelvis configuration (see also Appendix 7.2).

3.2.2. Level running at all speeds

A. Both in males and females **aged 6 to 13** (Figure 7.5), limits of each graph are: a) -30/30 mm along the forward direction; b) 710/860 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

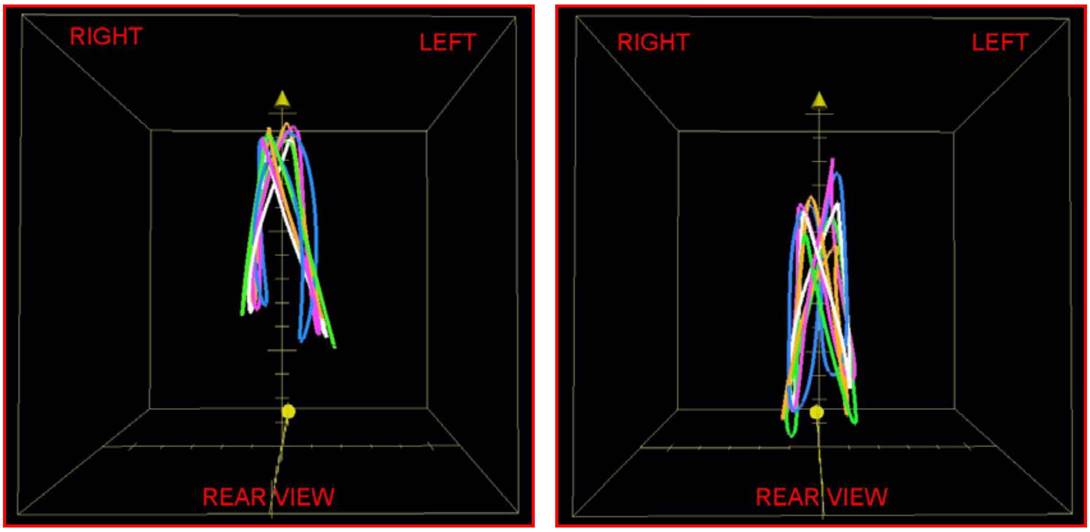


Figure 7.5. 3D *contours* as a function of all running speeds,
males (on the left) and females aged 6 to 13 (on the right).

On average, the BCOM is slightly raised and lifted in males because their higher heights compared to females. Similarly, in both males and females, the BCOM becomes slightly more vertical with increasing running speed.

B. Both in males and females **aged 25 to 35** (Figure 7.6), limits of each graph are: a) -30/30 mm along the forward direction; b) 910/1080 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

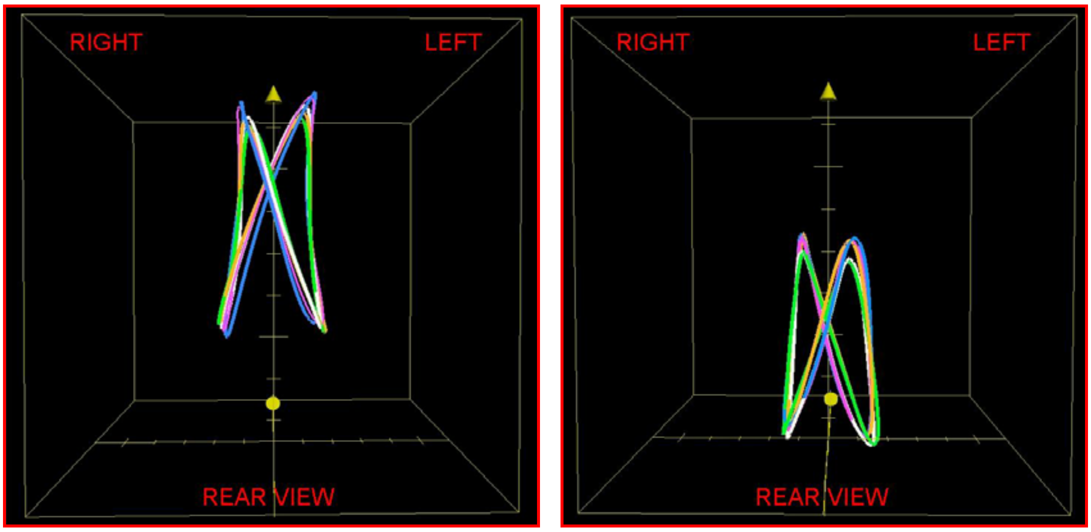


Figure 7.6. 3D *contours* as a function of all running speeds, males (on the left) and females aged 25 to 35 (on the right).

On average, the BCOM is slightly raised and lifted in males because their higher heights compared to females. Similarly, in both males and females, the BCOM becomes slightly more vertical with increasing running speed.

C. Both in males and females **aged 56 to 65** (Figure 7.7), limits of each graph are: a) -30/30 mm along the forward direction; b) 870/1050 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

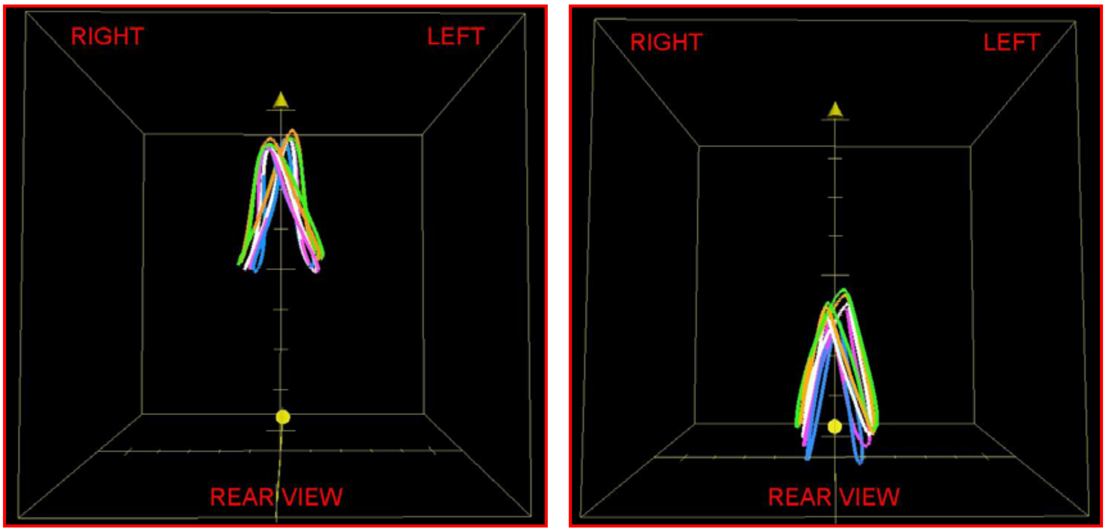


Figure 7.7. 3D *contours* as a function of all running speeds, males (on the left) and females aged 56 to 65 (on the right).

On average, the BCOM is slightly raised and lifted in males because their higher heights compared to females. Similarly, in both males and females, the BCOM becomes slightly more vertical with increasing running speed.

4. DIGITAL LOCOMOTORY SIGNATURE AS A FUNCTION OF AGE

4.1. Introduction

In the following sections, there are 3D *contours*, at the level gradient, in both males and females of each age group, in both walking and running (at single speeds). In each picture, we decided to insert all investigated age groups in order: **a)** to better visualize the absence of significant differences in subjects aged 14 to 55; and **b)** to emphasize the strange pattern of children and elderly adults in comparison to the other ages.

To be precise, a graphical top view of the average BCOM pattern has been proposed in each testing condition; moreover, the graphical limits in each direction have been defined.

4.2. Results of our experiments

DLS pattern in level walking at each speed has been represented as a function of age, as well.

A. In **males** (Figure 7.8, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 750/1020 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

At all the investigated speeds: **a)** children (white loop) have the most packed and irregular BCOM pattern. This becomes particularly evident as walking speed increases; **b)** there are not so visible significant differences in the other age groups although, in general, males aged 25 to 35 (yellow loop) seem to have the most regular *contour* independently of speed; furthermore, **c)** elderly adult males (pink loop) seem to have the most irregular and changing BCOM pattern. In addition, as a function of speed, the top view visualizes the progressive decreased lateral movement that implies a more vertical position of the BCOM.

Finally, in children there is a slightly asymmetry on the right side (Lund, 1930; Klimek et al., 2007) differently to what happens in all the other age groups. This is probably related to subjects being right-handed (Bracha et al., 1987; Delattre et al., 2001; Strike et al., 2009). Furthermore, it is important to remember that both structural and functional dominance (Sadeghi et al., 2000; Mohr et al., 2007; Souman et al., 2009) are related to: a) handedness; b) eyedness (Lund, 1930); c) length of arms (Lund, 1930); d) length of legs (Manello, 1992; Song et al., 1997; Strike et al., 2009); and e) posture (Lund, 1930; Gnat et al., 2009), as well.

B. In **females** (Figure 7.9, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 740/1020 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

As in males, at all the investigated speeds: **a)** children have the most packed and irregular BCOM pattern as walking speed increases; **b)** especially at the higher speeds, there are not so visible significant differences in the other age groups; in addition, **c)** elderly adult females seem to have the most changing BCOM pattern. In this case, there is the progressive decreased lateral movement, as well.

Finally, in children, at the lower speeds, there is a slightly asymmetry on the left side whereas, at the higher speeds, it turns on the right side. Uniformly, in all the other age groups the asymmetry is evident only on the left side and it is probably related to subjects being right-handed (Lund, 1930; Bracha et al., 1987; Delattre et al., 2001; Klimek et al., 2007).

DLS pattern in level running at each speed has been represented as a function of age, as well.

A. In **males** (Figure 7.10, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 760/1050 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

The top view visualizes the progressive increased vertical movement as a function of speed.

At all the investigated speeds: **a)** children have the most irregular BCOM pattern; **b)** there are only slightly visible differences in the other age groups: in general, males aged 25 to 35 seem to have the most regular *contour* independently of speed; finally, **c)** elderly adult males seem to have the most irregular BCOM pattern. Moreover, in all age groups, it becomes quite easy to distinguish the predominant asymmetry on the left side.

B. In **females** (Figure 7.11, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 740/1030 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

As already shown in males, at all the investigated speeds children and adults aged 56 to 65 have the most irregular BCOM pattern. Moreover, in all age groups, it becomes quite easy to distinguish the predominant asymmetry on the left side.

5. DIGITAL LOCOMOTORY SIGNATURE AS A FUNCTION OF SPEED

5.1. Introduction

As previously widely demonstrated, human locomotion (walking and running) could be speed-dependent. Of utmost importance, gait cycle variability changes with speed (Jordan et al., 2007; 2008). Therefore, in the following sections, there are 3D *contours*, at the level gradient, in both males and females aged 25 to 35, in both walking and running as a function of speed. We decided to insert only this age group because of its more regular pattern. However, all other age groups seem to behave quite similarly.

As shown in par. 4.1 above, a graphical rear view of the average BCOM pattern has been proposed in each testing condition; moreover, the graphical limits in each direction have been defined.

5.2. Results of our experiments

The next graphs contain the DLS pattern in level walking in subjects aged 25 to 35, as a function of speed.

A. In **males** (Figure 7.12 and 7.14), limits of each graph are (walking and running): a) -30/30 mm along the forward direction; b) 960/1080 mm along the vertical direction; and c) -30/30 mm along the lateral direction.

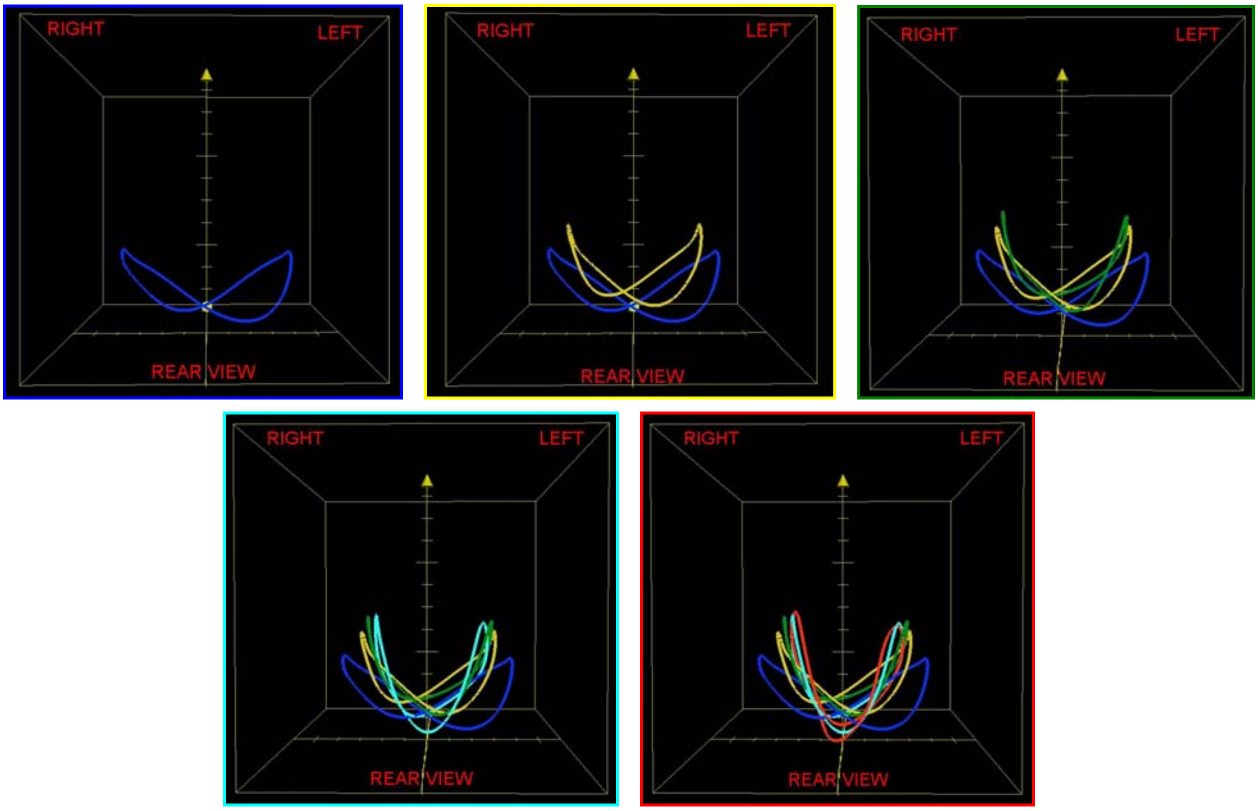
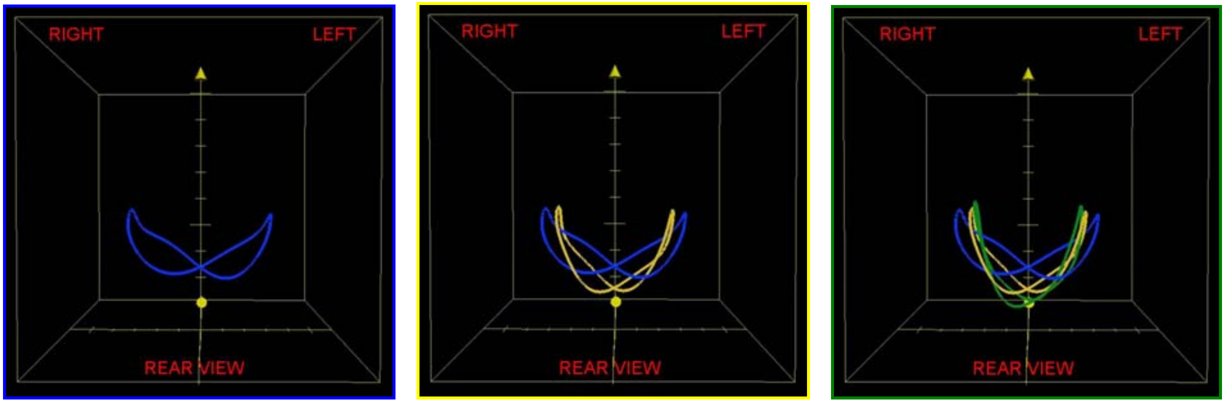


Figure 7.12. 3D *contours* in level walking at all speeds (from 0.83 to 1.94 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

B. In **females** (Figure 7.13 and 7.15), limits of each graph are (walking and running): a) -30/30 mm along the forward direction; b) 910/1010 mm along the vertical direction; and c) -30/30 mm along the lateral direction.



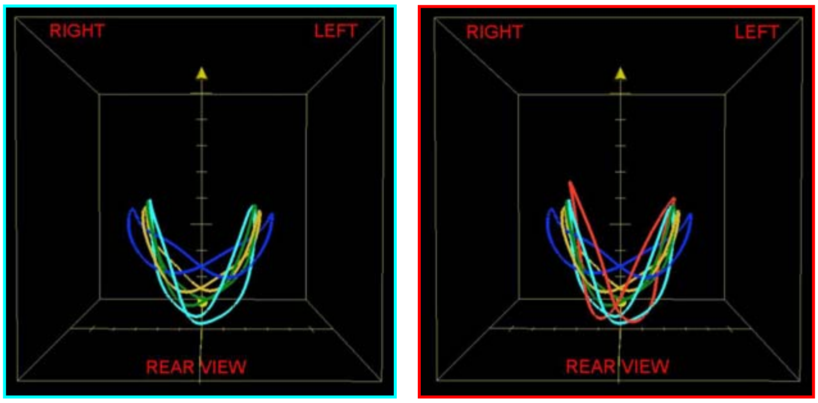


Figure 7.13. 3D *contours* in level walking at all speeds (from 0.83 to 1.94 m/s), females aged 25 to 35 (from the top left graph to the bottom right graph).

In level walking, the qualitative analysis of the average 3D *contours* demonstrates that:

1. males and females are quite similar, independently of age;
2. the lowest speeds (0.83 and 1.11 m/s) have the most peculiar DLS independently of age;
3. this pattern is related to the fact that these speeds are not so completely natural and common;
4. however, if walking speed increases, the 3D *contour* becomes more vertical and regular.

The next graphs contain the DLS pattern in level running in subjects aged 25 to 35, as a function of speed.

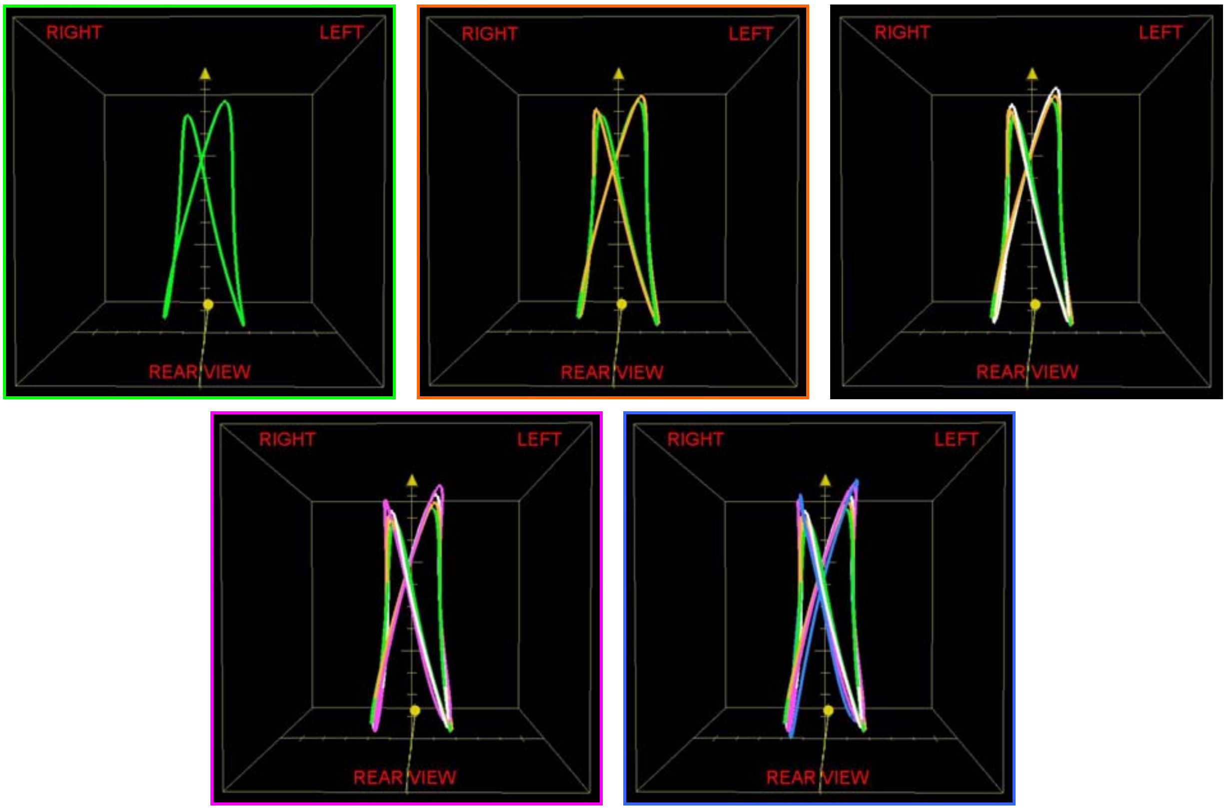


Figure 7.14. 3D *contours* in level running at all speeds (from 1.94 to 3.06 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

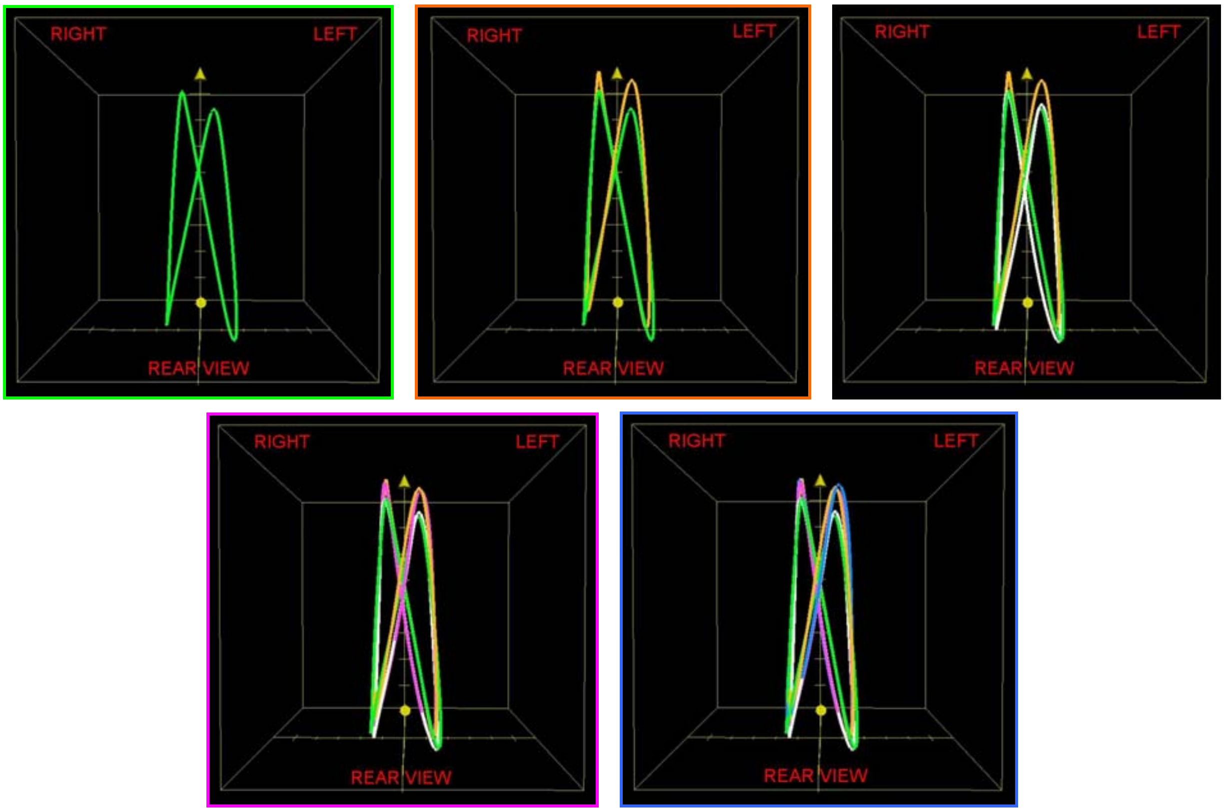


Figure 7.15. 3D *contours* in level running at all speeds (from 1.94 to 3.06 m/s), females aged 25 to 35 (from the top left graph to the bottom right graph).

In level running, the qualitative analysis of the average 3D *contours* demonstrates that:

1. males and females are quite similar, independently of age;
2. the lowest speeds (1.94 and 2.22 m/s) have the most peculiar DLS independently of age;
3. this pattern is related to the fact that these speeds are only slightly natural and common;
4. however, if running speed increases, the 3D *contour* becomes more vertical and regular.

6. DIGITAL LOCOMOTORY SIGNATURE AS A FUNCTION OF GRADIENT

6.1. Introduction

In the following sections, there are 3D *contours*, at different slopes (both downhill and uphill), in males aged 25 to 35, in both walking and running (at each speed). We decided to insert only male graphs because of the high similarity between males and females, in each gradient condition. Specifically, the vertical position of the BCOM (Ay0) has been obtained by sub-tracking the height of the treadmill which has been empirically measured: indeed, it measures 51 cm from the ground (see also in the enclosed CD, in First Study, Chapter 7, file Treadmill dimensions).

In this case, because of the high number and complexity of testing conditions, only a qualitative analysis (see also par. 2.1 above) has been performed. As shown in par. 4.1 and 5.1 above, a graphical rear view of the average BCOM pattern has been proposed in each testing condition; moreover, the graphical limits in each direction have been defined.

6.2. Results of our experiments

6.2.1. Gradient walking

A. In **downhill gradient walking** (Figure 7.16, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 960/1300 mm along the vertical direction; and c) -30/30 mm along the lateral direction. Our results show that:

- similarly to level gradient, the lower speeds seem to have the most strange and unusual pattern;
- furthermore, similarly to level gait, the BCOM pattern becomes more vertical as a function of speed, independently of downhill gradient;
- as expected, due to an increasing height of the treadmill, the BCOM lifts and accelerates as a function of downhill gradient. However, this behaviour is more evident at the lowest slopes. Probably, such a pattern could be related to an increased metabolic requirement instead of a major motor control (Minetti et al., 1993; 1994; Jordan et al., 2007).

B. In **uphill gradient walking** (Figure 7.17, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 960/1480 mm along the vertical direction; and c) -30/30 mm along the lateral direction. Our results show that:

- similarly to level gradient, the lower speeds seem to have the most strange and unusual pattern;
- furthermore, similarly to level gait, the BCOM pattern becomes more vertical as a function of speed, independently of uphill gradient;
- as expected, due to an increasing height of the treadmill, the BCOM lifts and accelerates as a function of uphill gradient. This behaviour is similar at all slopes. As discussed above, such a pattern could be related to an increased metabolic requirement instead of a major motor control (Minetti et al., 1993; 1994; Jordan et al., 2007).

6.2.2. Gradient running

A. In **downhill gradient running** (Figure 7.18, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 960/1360 mm along the vertical direction; and c) -30/30 mm along the lateral direction. Our results show that:

- similarly to level gradient, the lower speeds seem to have the most unusual pattern;

- furthermore, similarly to level gait, the BCOM pattern becomes more vertical as a function of speed, independently of downhill gradient;
- as expected, due to an increasing height of the treadmill, the BCOM lifts and accelerates as a function of downhill gradient. This behaviour is similar at all slopes (i.e. the increased metabolic requirement instead of the bigger motor control).

B. In **uphill gradient running** (Figure 7.19, at the end of the chapter), limits of each graph are: a) -30/30 mm along the forward direction; b) 960/1460 mm along the vertical direction; and c) -30/30 mm along the lateral direction. Our results show that:

- similarly to level gradient, the lower speeds seem to have the most unusual pattern;
- furthermore, similarly to level gait, the BCOM pattern becomes more vertical as a function of speed, independently of uphill gradient;
- as expected, due to an increasing height of the treadmill, the BCOM lifts and accelerates as a function of uphill gradient. This behaviour is similar at all slopes (i.e. the increased metabolic requirement instead of the bigger motor control).

7. DISCUSSION AND CONCLUSION

By using amplitude and phase (symmetrical and asymmetrical) coefficients, Digital Locomotory Signature has been mathematically calculated as gender, age, type of locomotion, speed and gradient change. Clearly, a closed loop (dependent on the variable 'time') following the same pattern at each stride is then obtained.

The analysis of this global index of the BCOM dynamics reveals that:

a) on average, male *contours* are slightly higher (along vertical direction) than female *contours* because of male higher anthropometric dimensions (i.e. height);

b) as expected, three main age groups could be single out: children aged 6 to 13, young adults aged 25 to 35 and elderly adults aged 56 to 65. Moreover, other age groups (ranging from 14 to 55 years) are quite similar to young adults. Therefore, only these three groups have been considered in statistical analyses;

c) independently of type of locomotion and speed, children aged 6 to 13 and elderly adults aged 56 to 65 have the most changing, strange and unusual pattern. According to literature, this pattern is more pronounced at the lowest speeds;

d) independently of gait condition, human locomotion is rather asymmetrical. By deeply analysing 3D *contours*, it makes clear that the global asymmetry is more marked on the left side. This is probably related to subjects being right-handed: indeed, 95% of studied people was predominant right-hand;

e) to be precise, this tendency has been suggested to be mediated by hemispherical asymmetries in the dopamine system (according to Bracha et al., 1987; Fitzpatrick et al., 1999; Mohr et al., 2003; 2004; 2007) or by caloric and galvanic stimulation (Marques et al., 2007). However, an alternative explanation focuses on biomechanical asymmetries (Souman et al., 1999), such as differences in leg length or leg strength (Guldberg, 1897; Lund, 1930; Gurney, 2002; Gnat et al., 2009). On this account, most humans would have one leg longer or stronger than the other, creating a small but constant bias in the opposite direction (Souman et al., 2009). Clearly, this occurs in the most people;

f) independently of age, the lowest walking and running speeds seem to be not so completely natural and common. Hypothetically, they seem to be too slow. As a consequence, if lower speeds will be investigated in the near future, we will expect a more strange and no-regular three-dimensional displacement of the BCOM;

g) the BCOM lifts and accelerates becoming more vertical as a function of both walking and running speed (Lee et al., 2009);

h) however, the variable gradient seems not to play a key role.

Positively, all these important results wholly characterize in a novel way the 3D trajectories of the body centre of mass in healthy humans during locomotion in a wide number of testing conditions.

Therefore, the initial *database* of reference equation coefficients has been successfully built up.

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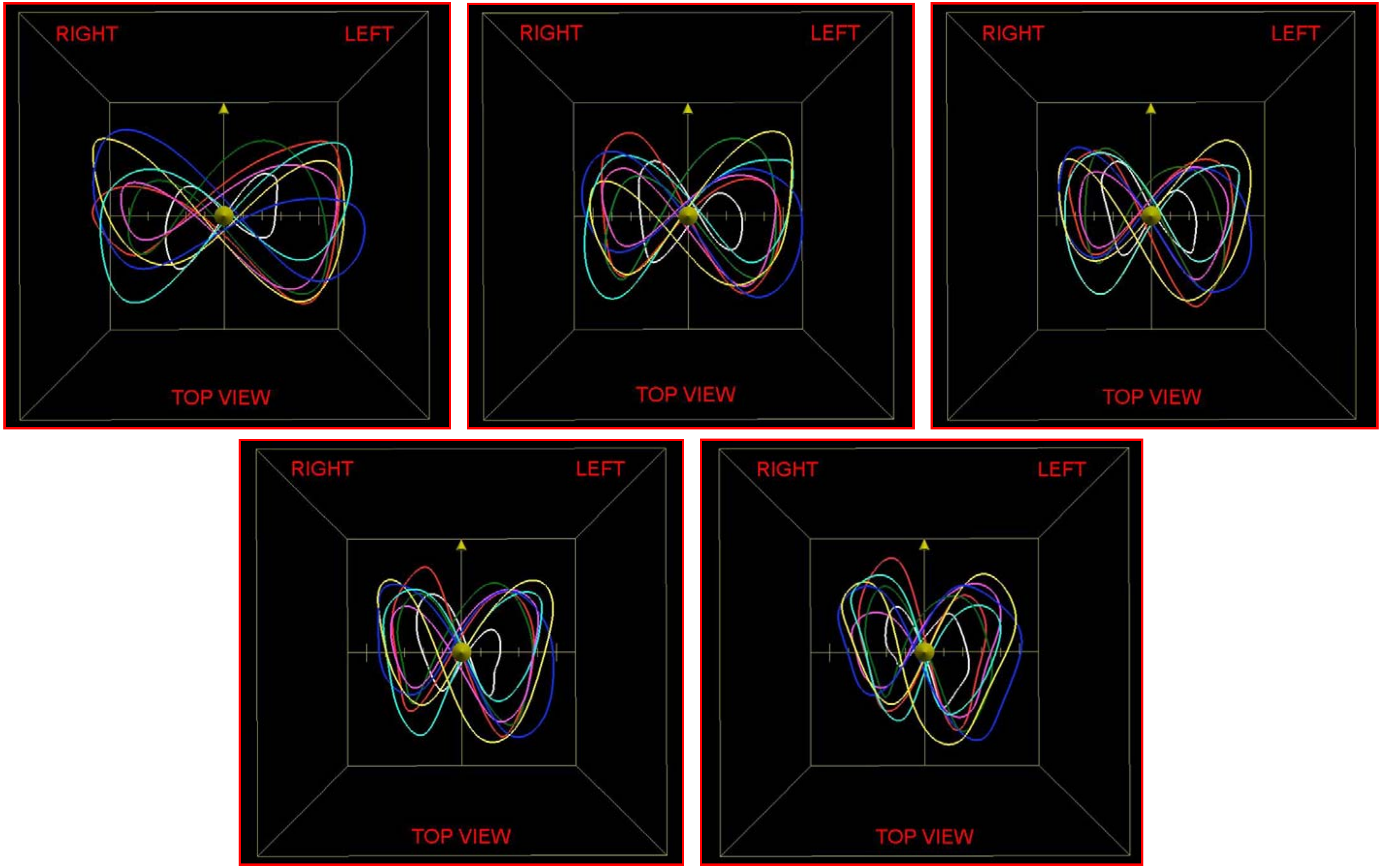

Figure 7.8. 3D *contours* as a function of age in males, level walking from 0.83 to 1.94 m/s (from the top left graph to the bottom right graph).

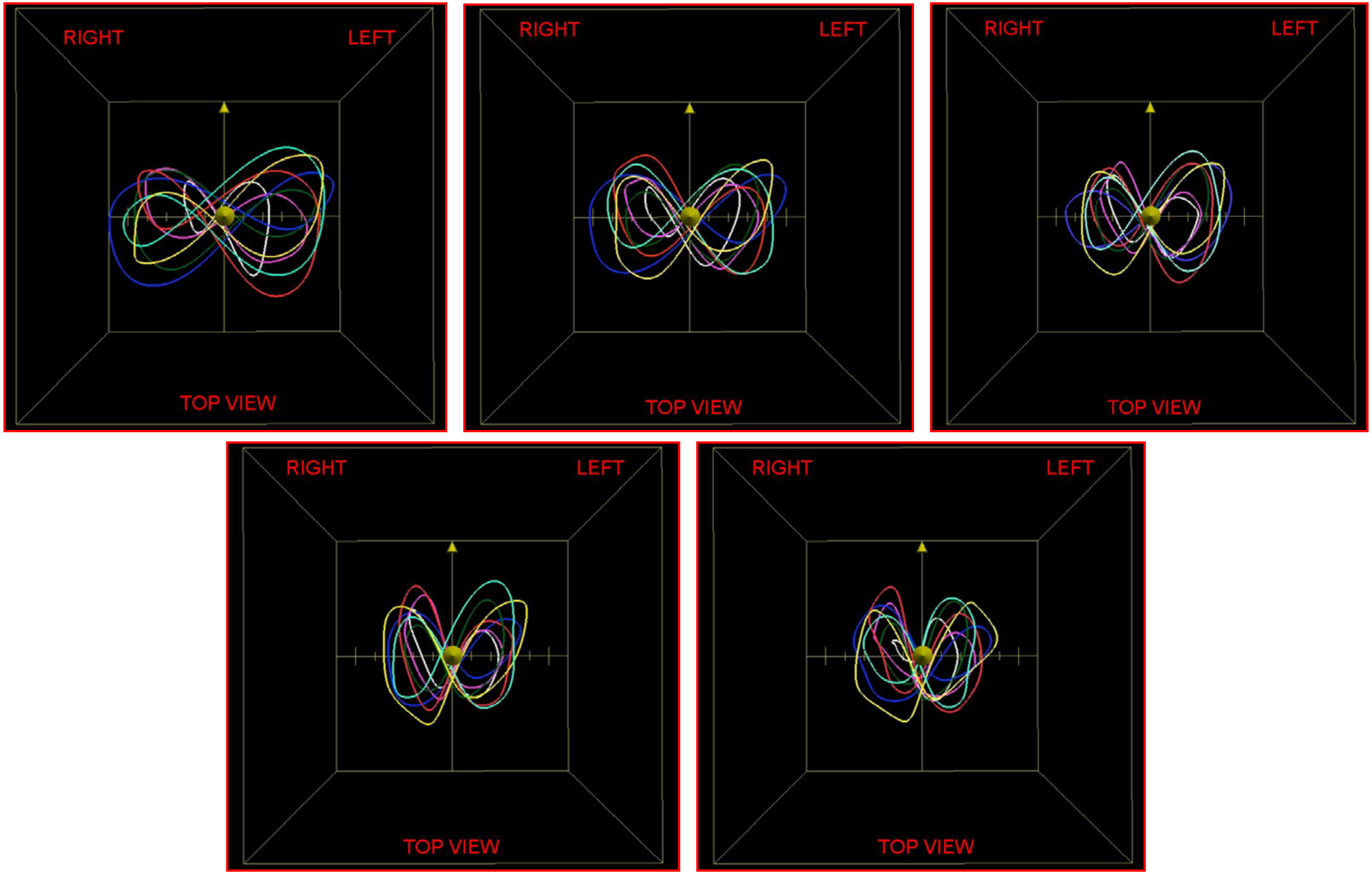

Figure 7.9. 3D *contours* as a function of age in females, level walking from 0.83 to 1.94 m/s (from the top left graph to the bottom right graph).

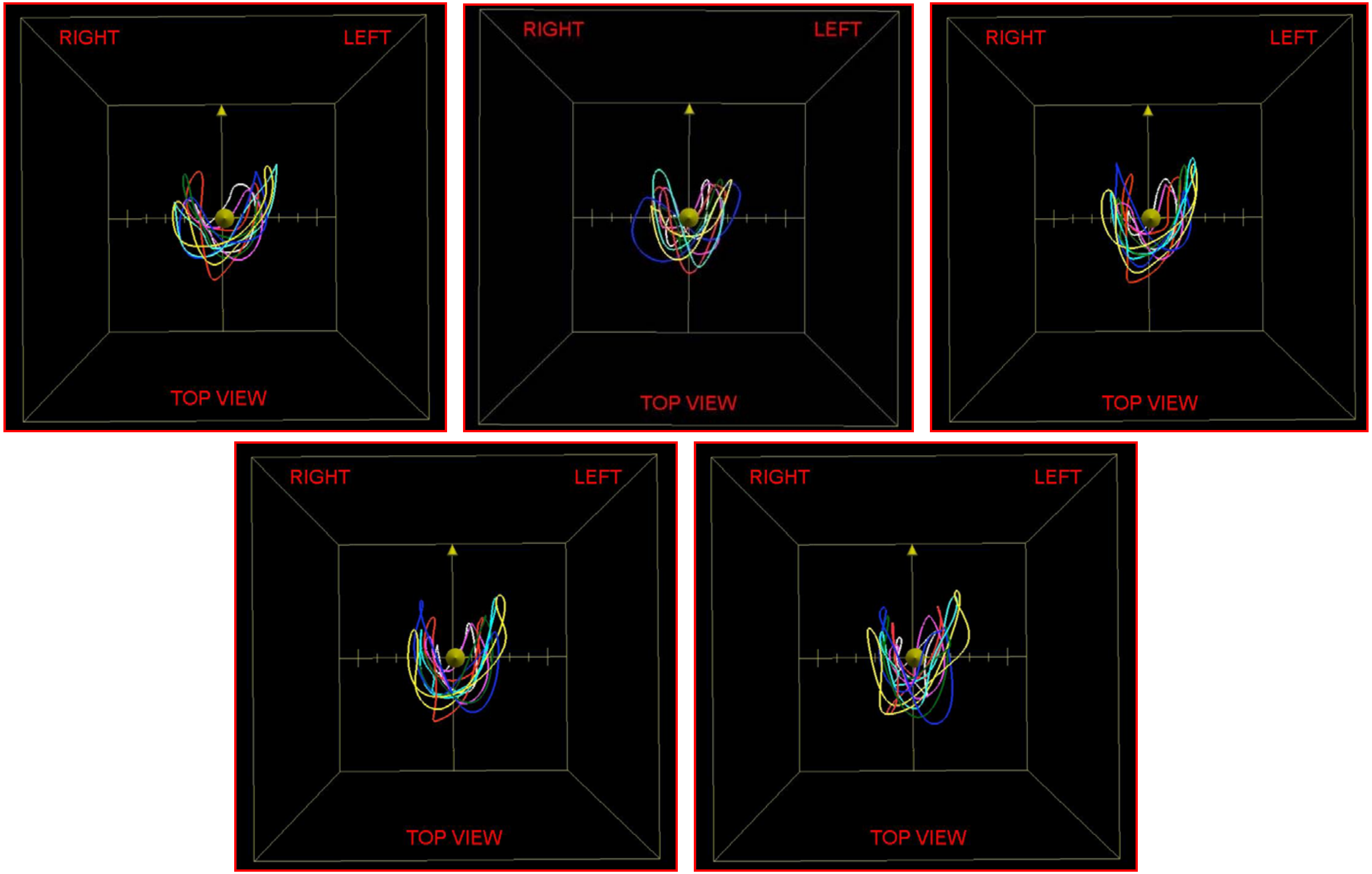

Figure 7.10. 3D *contours* as a function of age in males, level running from 1.94 to 3.06 m/s (from the top left graph to the bottom right graph).

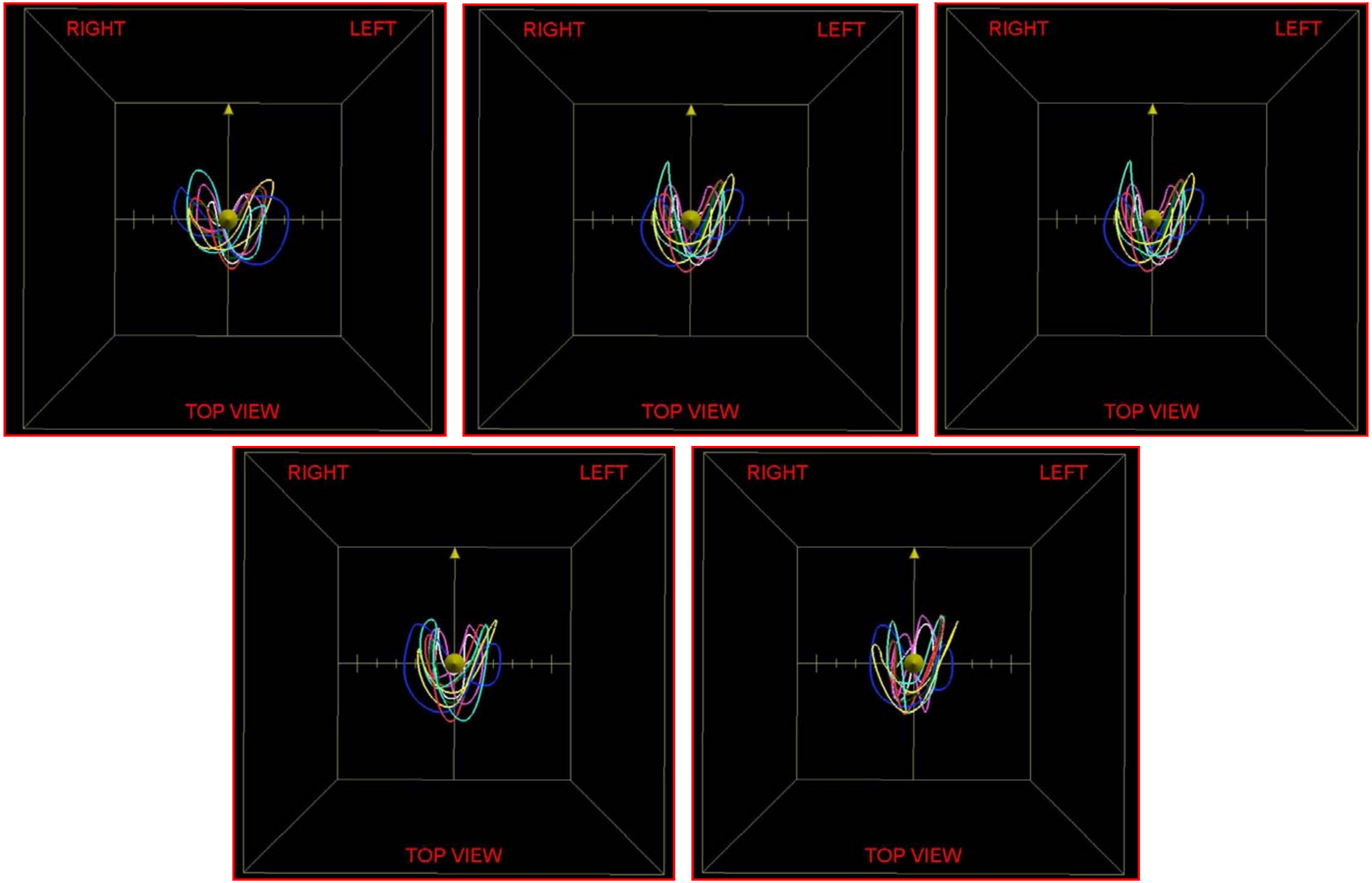

Figure 7.11. 3D *contours* as a function of age in females, level running from 1.94 to 3.06 m/s (from the top left graph to the bottom right graph).

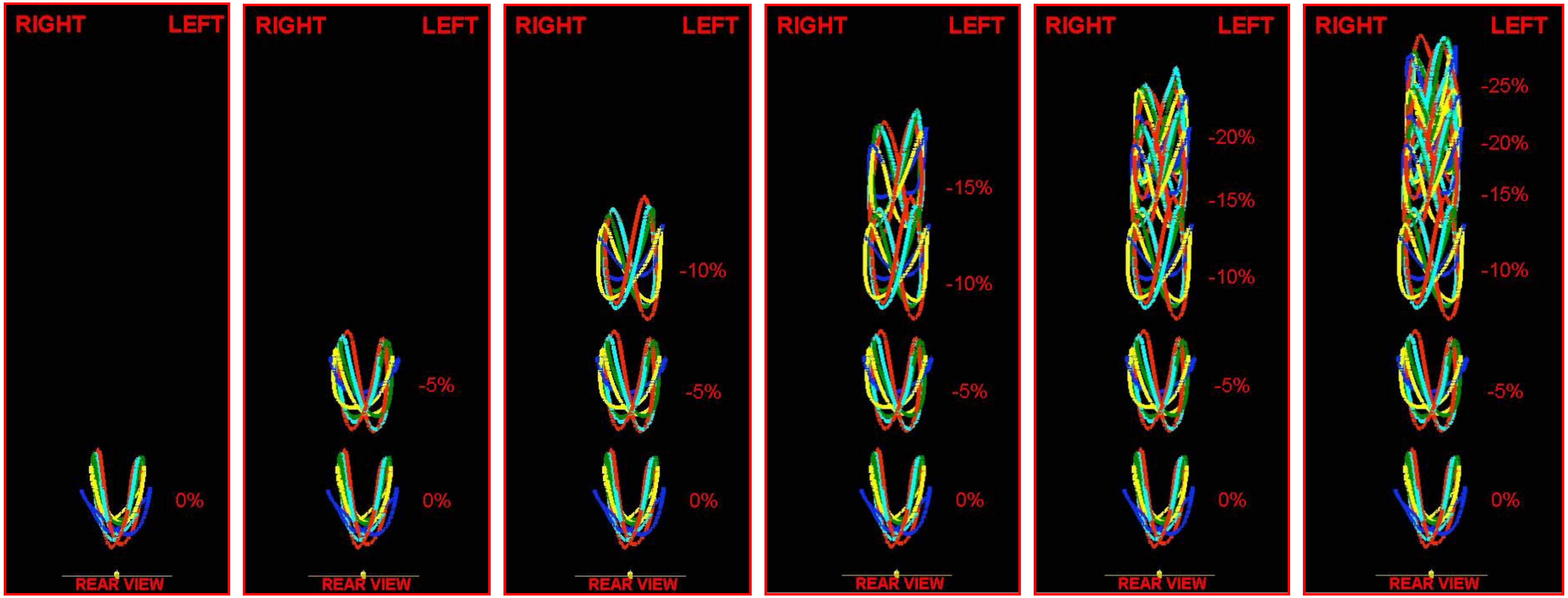

Figure 7.16. 3D *contours* in downhill gradient (from 0% to -25%) walking at all speeds (from 0.83 to 1.94 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

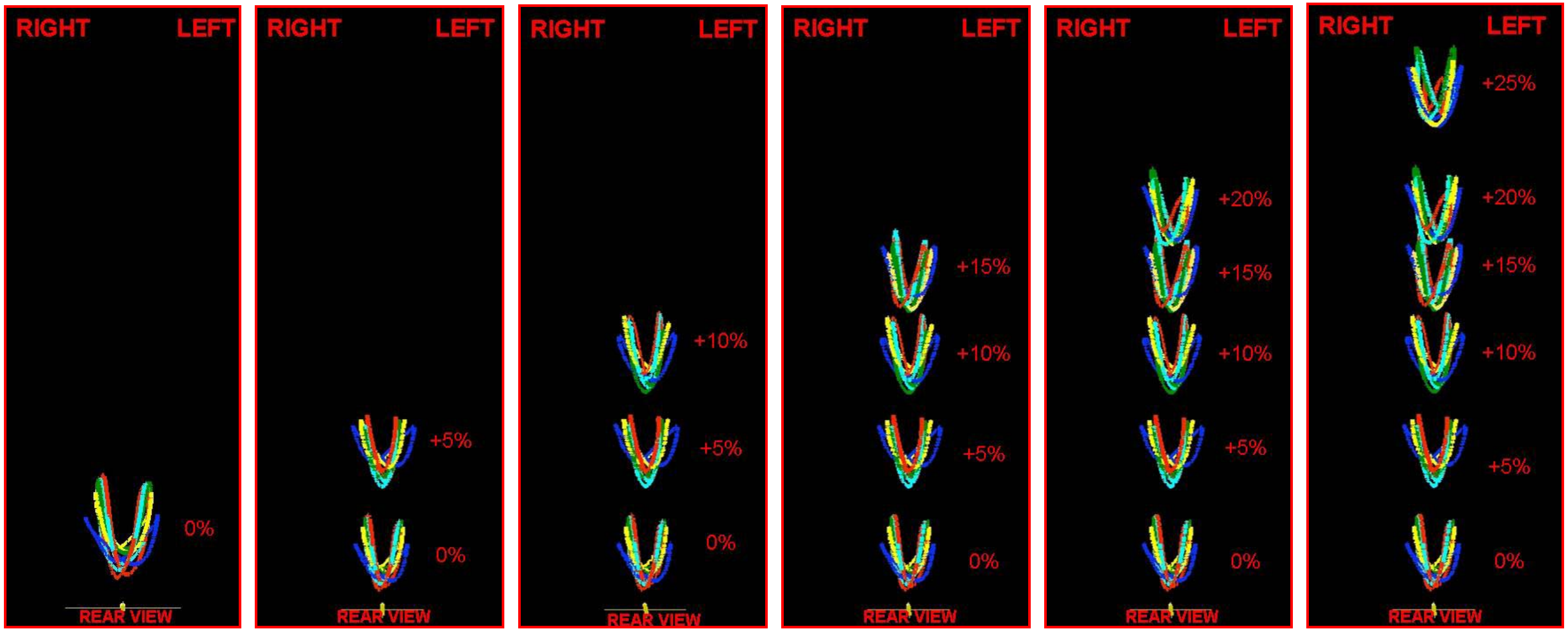

Figure 7.17. 3D *contours* in uphill gradient (from 0% to 25%) walking at all speeds (from 0.83 to 1.94 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

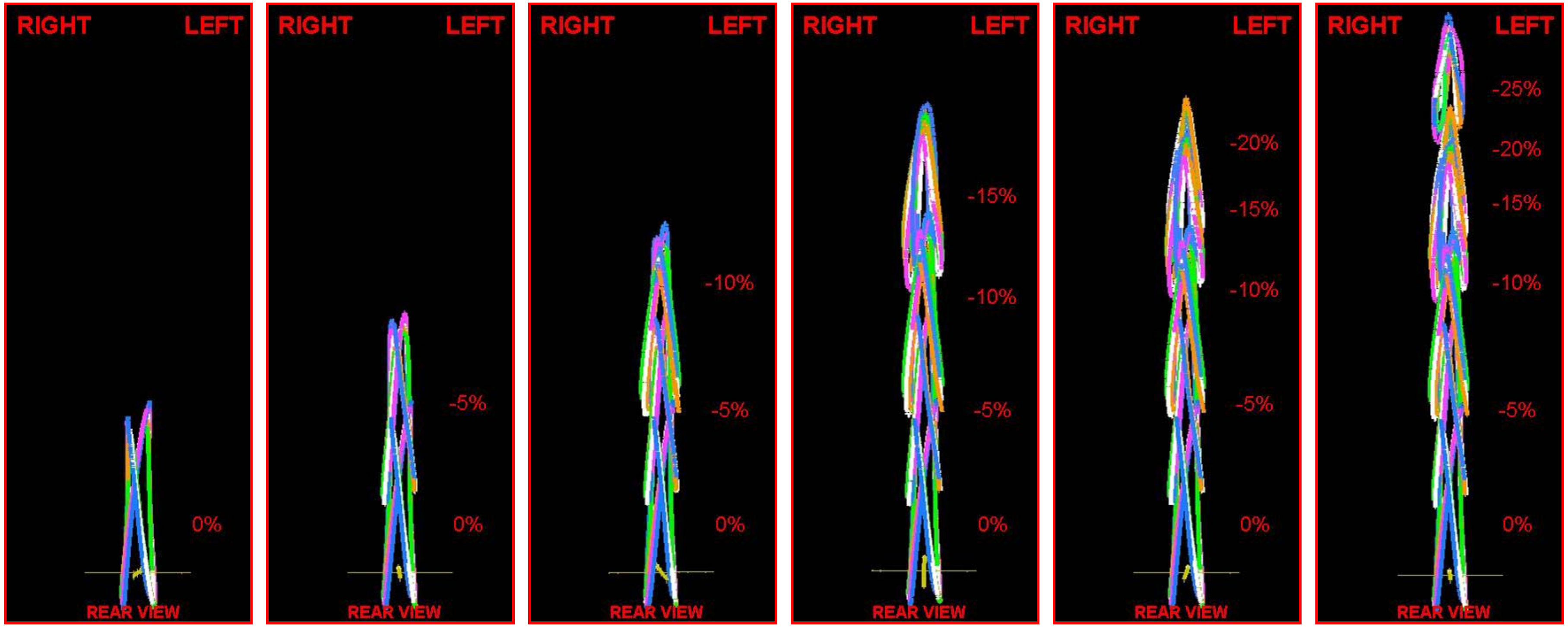

Figure 7.18. 3D *contours* in downhill gradient (from 0% to -25%) running at all speeds (from 1.94 to 3.06 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

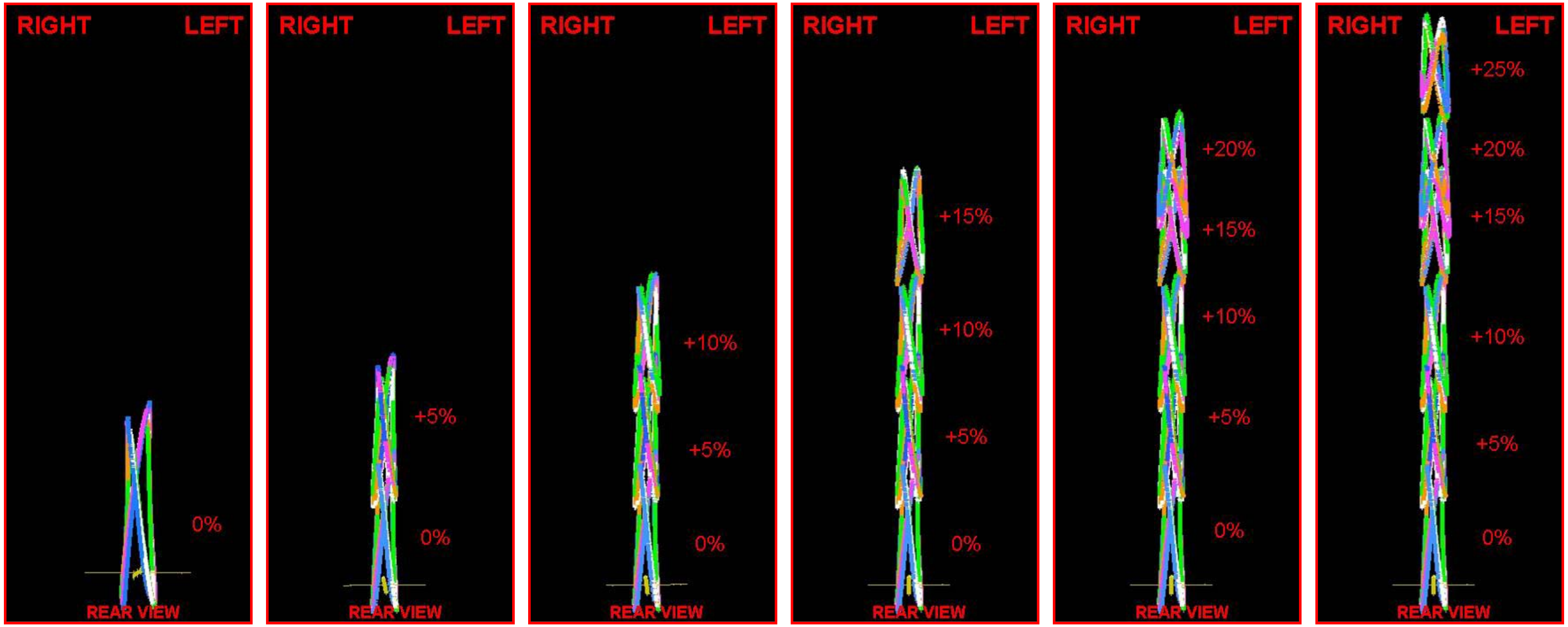

Figure 7.19. 3D *contours* in uphill gradient (from 0% to 25%) running at all speeds (from 1.94 to 3.06 m/s), males aged 25 to 35 (from the top left graph to the bottom right graph).

Appendix 7.1

HARMONIC COEFFICIENTS: STATISTICAL ANALYSIS

1. GENDER, AGE AND SPEED: RESULTS OF STATISTICAL ANALYSIS

As previously described in chapter 7 (par. 2.2), a one-way ANOVA with Huynh-Feldt correction for repeated measures were carried out to assess the main differences in amplitude coefficients (A1, A2, A3 and A4) and sine and cosine functions (derived from φ1, φ2, φ3 and φ4) between males and females as walking/running speed changes, in children aged 6 to 13, young adults aged 25 to 35 and elderly adults aged 56 to 65.

However, it is important to remember that such a statistical analysis is not fully completed regarding phase coefficients. Indeed, they represent circular variables upon which a proper circular statistical approach has to be applied. Therefore, the results we proposed are partially available primarily because of the high deviation to Gaussian distribution that could be often observed in these coefficients (as stated for other measurements ranging from -1 to 1). In any case, we think that our results could constitute an important start point for futher and deeper investigations.

Therefore, the tables in the following paragraphs contain the results of such an analysis.

2. LEVEL WALKING

2.1. Level walking in subjects aged 6 to 13

A. In the forward direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ax2, there is a little significance as walking speed increases ($p<0.05$); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine and cosine functions, no significances are found both as a function of walking speed and in the interaction between gender and speed.

VARIABLE	**Ax1**	**Ax2**	**Ax3**	**Ax4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.03 **(p<0.05)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Sinx1**	**Sinx2**	**Sinx3**	**Sinx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosx1	Cosx2	Cosx3	Cosx4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.1a. Amplitude coefficients and sine/cosine functions

in the forward walking direction (subjects aged 6 to 13).

B. In the vertical direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ay1 and Ay2, there is a high significance as walking speed increases ($p<0.001$); therefore, c) in the previous amplitudes, the interaction between gender and speed is slightly significant, as well ($p<0.05$). Furthermore, d) in siney1 and siney2, there is a slightly significance as walking speed increases ($p<0.05$ and $p<0.01$, respectively); e) in cosiney3, there is a low significance, as well ($p<0.05$); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Ay1	Ay2	Ay3	Ay4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	<0.0001 **(p<0.001)**	<0.0001 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	0.04 **(p<0.05)**	0.02 **(p<0.05)**	p=NS	p=NS

VARIABLE	Siny1	Siny2	Siny3	Siny4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	0.02 **(p<0.05)**	0.003 **(p<0.01)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	0.04 **(p<0.05)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.1b. Amplitude coefficients and sine/cosine functions

in the vertical walking direction (subjects aged 6 to 13).

C. In the lateral direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Az1 and Az4, there is a significance as walking speed increases ($p<0.001$ and $p<0.01$, respectively); therefore, c) in Az1, the interaction between gender and speed is slightly significant, as well ($p<0.05$). Furthermore, d) sinez1 and cosinez1 are not involved in the statistical analysis because of their constant value (for more details see chapter 6, par. 5.2.2); e) in cosinez3, there is a low significance as a function of walking speed ($p<0.05$); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	**Az1**	**Az2**	**Az3**	**Az4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	0.0002 **(p<0.001)**	p=NS	p=NS	0.005 **(p<0.01)**
Interaction Gender/Speed	0.02 **(p<0.05)**	p=NS	p=NS	p=NS

VARIABLE	**Sinz2**	**Sinz3**	**Sinz4**
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

VARIABLE	**Cosz2**	**Cosz3**	**Cosz4**
Gender	p=NS	p=NS	p=NS
Speed	p=NS	0.01 **(p<0.05)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

Table 7.1c. Amplitude coefficients and sine/cosine functions
in the lateral walking direction (subjects aged 6 to 13).

2.2. Level walking in subjects aged 25 to 35

A. In the forward direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in all amplitudes, there is a significance as walking speed increases; however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in cosinex4, there is a significance as a function of walking speed (p<0.01); however, e) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	**Ax1**	**Ax2**	**Ax3**	**Ax4**
Gender	p=NS	0.02	p=NS	p=NS
Speed	0.004 **(p<0.01)**	0.04 **(p<0.05)**	0.001 **(p<0.01)**	0.0002 **(p<0.001)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Sinx1**	**Sinx2**	**Sinx3**	**Sinx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Cosx1**	**Cosx2**	**Cosx3**	**Cosx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	0.009 **(p<0.01)**

Table 7.2a. Amplitude coefficients and sine/cosine functions
in the forward walking direction (subjects aged 25 to 35).

B. In the vertical direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ay1 and Ay2, there is a high significance as walking speed increases (p<0.001); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in siney4 and cosiney2, there is a significance as a function of walking speed ($p<0.01$ and $p<0.001$, respectively); however, e) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Ay1	Ay2	Ay3	Ay4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	0.0009 **(p<0.001)**	<0.0001 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Siny1	Siny2	Siny3	Siny4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.001 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.0001 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.2b. Amplitude coefficients and sine/cosine functions
in the vertical walking direction (subjects aged 25 to 35).

C. In the lateral direction, our results show that: a) in Az2 and Az4, males and females slightly differ; b) in Az1 and Az3, there is a significance, as walking speed increases ($p<0.001$ and $p<0.01$, respectively); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine/cosine functions, males and females are similar; e) in cosinez3, there is a high significance as a function of walking speed ($p<0.01$); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Az1	Az2	Az3	Az4
Gender	p=NS	0.007 **(p<0.01)**	p=NS	0.02 **(p<0.05)**
Speed	0.0001 **(p<0.001)**	p=NS	0.004 **(p<0.01)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinz2	Sinz3	Sinz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

VARIABLE	Cosz2	Cosz3	Cosz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	0.002 **(p<0.01)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

Table 7.2c. Amplitude coefficients and sine/cosine functions
in the lateral walking direction (subjects aged 25 to 35).

2.3. Level walking in subjects aged 56 to 65

A. In the forward direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in all amplitudes, there is a significance as walking speed increases; however, c) the interaction between gender and speed is not significant. Furthermore, d) in sinex4, there is a significance as a function of walking speed ($p<0.01$); in addition, e) in cosinex2 and cosinex4, there is a significance as a function of walking speed ($p<0.05$ and $p<0.001$, respectively); however, f) only in cosinex4 functions, the interaction between gender and speed is slight significant.

VARIABLE	Ax1	Ax2	Ax3	Ax4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	0.0009 **(p<0.001)**	0.03 **(p<0.05)**	0.04 **(p<0.05)**	<0.0001 **(p<0.001)**
Interaction Gender/Speed	p=NS	0.02	p=NS	p=NS

VARIABLE	Sinx1	Sinx2	Sinx3	Sinx4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.001 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosx1	Cosx2	Cosx3	Cosx4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.013 **(p<0.05)**	p=NS	0.0002 **(p<0.001)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	0.009 **(p<0.01)**

Table 7.3a. Amplitude coefficients and sine/cosine functions
in the forward walking direction (subjects aged 56 to 65).

B. In the vertical direction, our results show that: a) in Ay2, males and females slightly differ ($p<0.05$); b) in Ay2, there is a high significance as walking speed increases ($p<0.001$); therefore, c) in Ay2, the interaction between gender and speed is significant, as well ($p<0.01$). Furthermore, d) in all sine/cosine functions, males and females are similar; e) in siney4 and cosiney2, there is a significance as a function of walking speed ($p<0.05$ and $p<0.001$, respectively); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Ay1	Ay2	Ay3	Ay4
Gender	p=NS	0.02 **(p<0.05)**	p=NS	p=NS
Speed	p=NS	<0.0001 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.001 **(p<0.01)**	p=NS	p=NS

VARIABLE	Siny1	Siny2	Siny3	Siny4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.03 **(p<0.05)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	<0.0000 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.3b. Amplitude coefficients and sine/cosine functions
in the vertical walking direction (subjects aged 56 to 65).

C. In the lateral direction, our results show that: a) in Az3, males and females slightly differ (p<0.05); b) in Az1 and Az4, there is a significance, as walking speed increases (p<0.001 and p<0.01, respectively); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine/cosine functions, males and females are similar; e) in sinez3 and cosinez3, there is a significance as a function of walking speed (p<0.01 and p<0.05, respectively); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Az1	Az2	Az3	Az4
Gender	p=NS	p=NS	0.01 **(p<0.05)**	p=NS
Speed	<0.0001 **(p<0.001)**	p=NS	p=NS	0.002 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinz2	Sinz3	Sinz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.006 **(p<0.01)**	p=NS

VARIABLE	Cosz2	Cosz3	Cosz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	0.01 **(p<0.05)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

Table 7.3c. Amplitude coefficients and sine/cosine functions
in the lateral walking direction (subjects aged 56 to 65).

3. LEVEL RUNNING

3.1. Level running in subjects aged 6 to 13

A. In the forward direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ax4, there is a significance, as running speed increases (p<0.01); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in sinex2 and cosinex2, there is a high significance as a function of walking speed (p<0.01 and p<0.001, respectively); therefore; e) the interaction between gender and speed is significant in these functions, as well (p<0.001 and p<0.01, respectively).

VARIABLE	Ax1	Ax2	Ax3	Ax4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.001 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinx1	Sinx2	Sinx3	Sinx4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.003 **(p<0.01)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.0006 **(p<0.001)**	p=NS	p=NS

VARIABLE	Cosx1	Cosx2	Cosx3	Cosx4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.0006 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.002 **(p<0.01)**	p=NS	p=NS

Table 7.4a. Amplitude coefficients and sine/cosine functions
in the forward running direction (subjects aged 6 to 13).

B. In the vertical direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in all amplitudes, there is no significant change as a function of running speed; therefore, c) the interaction between gender and speed is not significant. Furthermore, d) in cosiney2, there is a low significance as a function of running speed (p<0.05); therefore; e) in cosiney2 and cosiney3, the interaction between gender and speed is significant, as well (p<0.01 and p<0.05, respectively).

VARIABLE	Ay1	Ay2	Ay3	Ay4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Siny1	Siny2	Siny3	Siny4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.03 **(p<0.05)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.003 **(p<0.01)**	0.03 **(p<0.05)**	p=NS

Table 7.4b. Amplitude coefficients and sine/cosine functions
in the vertical running direction (subjects aged 6 to 13).

C. In the lateral direction, our results show that: a) in Az4, males and females differ (p<0.01); b) in Az1, there is a little significance as running speed increases (p<0.05); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine/cosine functions, males and females are similar; finally, e) in cosinez3, there is a low significance in the interaction between gender and speed (p<0.05).

VARIABLE	**Az1**	**Az2**	**Az3**	**Az4**
Gender	p=NS	p=NS	p=NS	0.003 **(p<0.01)**
Speed	0.04 **(p<0.05)**	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Sinz2**	**Sinz3**	**Sinz4**
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

VARIABLE	**Cosz2**	**Cosz3**	**Cosz4**
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.01 **(p<0.05)**	p=NS

Table 7.4c. Amplitude coefficients and sine/cosine functions
in the lateral running direction (subjects aged 6 to 13).

3.2. Level running in subjects aged 25 to 35

A. In the forward direction, our results show that: a) in Ax2, males and females slightly differ (p<0.05); b) in Ax2 and Ax4, there is a high significance as running speed increases (p<0.001); therefore, c) in Ax1, the interaction between gender and speed is slightly significant (p<0.05). Furthermore, d) in all sine/cosine functions, males and females are similar; e) in sinex4, there is a low significance as a function of running speed (p<0.05); finally, f) in cosinex3, the interaction between gender and speed is slightly significant (p<0.05).

VARIABLE	**Ax1**	**Ax2**	**Ax3**	**Ax4**
Gender	p=NS	0.04 **(p<0.05)**	p=NS	p=NS
Speed	p=NS	0.0001 **(p<0.001)**	p=NS	<0.0001 **(p<0.001)**
Interaction Gender/Speed	0.04 **(p<0.05)**	p=NS	p=NS	p=NS

VARIABLE	**Sinx1**	**Sinx2**	**Sinx3**	**Sinx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.02 **(p<0.05)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Cosx1**	**Cosx2**	**Cosx3**	**Cosx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	0.01 **(p<0.05)**	p=NS

Table 7.5a. Amplitude coefficients and sine/cosine functions
in the forward running direction (subjects aged 25 to 35).

B. In the vertical direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ay2 and Ay4, there is a significance as running speed increases (p<0.05 and p<0.01, respectively); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in siney2 and siney4, there is a significance as a function of running speed ($p<0.05$ and $p<0.001$, respectively); however, e) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Ay1	Ay2	Ay3	Ay4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.02 **(p<0.05)**	p=NS	0.007 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Siny1	Siny2	Siny3	Siny4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.04 **(p<0.05)**	p=NS	0.0007 **(p<0.001)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.5b. Amplitude coefficients and sine/cosine functions in the vertical running direction (subjects aged 25 to 35).

C. In the lateral direction, our results show that: a) in Az3 and Az4, males and females slightly differ ($p<0.05$); b) in Az4, there is a little significance as running speed increases ($p<0.05$); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine/cosine functions, males and females are similar; e) in sinez3, there is a low significance as a function of running speed ($p<0.05$); however, f) in all sine and cosine functions, the interaction between gender and speed is not significant.

VARIABLE	Az1	Az2	Az3	Az4
Gender	p=NS	p=NS	0.04 **(p<0.05)**	0.02 **(p<0.05)**
Speed	p=NS	p=NS	p=NS	0.04 **(p<0.05)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinz2	Sinz3	Sinz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	0.04 **(p<0.05)**	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

VARIABLE	Cosz2	Cosz3	Cosz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

Table 7.5c. Amplitude coefficients and sine/cosine functions in the lateral running direction (subjects aged 25 to 35).

3.3. Level running in subjects aged 56 to 65

A. In the forward direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Ax2, there is a significance, as running speed increases ($p<0.01$); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in sinex2, there is a high significance as a function of running speed ($p<0.001$); therefore, e) in this sine function, the interaction between gender and speed is significant, as well ($p<0.01$).

VARIABLE	**Ax1**	**Ax2**	**Ax3**	**Ax4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.005 **(p<0.01)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Sinx1**	**Sinx2**	**Sinx3**	**Sinx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.0001 **(p<0.001)**	p=NS	p=NS
Interaction Gender/Speed	p=NS	0.005 **(p<0.01)**	p=NS	p=NS

VARIABLE	**Cosx1**	**Cosx2**	**Cosx3**	**Cosx4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.6a. Amplitude coefficients and sine/cosine functions
in the forward running direction (subjects aged 56 to 65).

B. In the vertical direction, our results show that: a) in Ay4, males and females slightly differ ($p<0.05$); b) in all amplitudes, there is not a significance as a function of running speed; therefore, c) the interaction between gender and speed is not significant. Furthermore, d) in all sine/cosine functions, males and females are similar; e) in siney4, there is a high significance as a function of running speed ($p<0.01$); therefore, f) in this sine function, the interaction between gender and speed is slightly significant, as well ($p<0.05$).

VARIABLE	**Ay1**	**Ay2**	**Ay3**	**Ay4**
Gender	p=NS	p=NS	p=NS	0.02 **(p<0.05)**
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	**Siny1**	**Siny2**	**Siny3**	**Siny4**
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	0.008 **(p<0.01)**
Interaction Gender/Speed	p=NS	p=NS	p=NS	0.02 **(p<0.05)**

VARIABLE	Cosy1	Cosy2	Cosy3	Cosy4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

Table 7.6b. Amplitude coefficients and sine/cosine functions in the vertical running direction (subjects aged 56 to 65).

C. In the lateral direction, our results show that: a) in all amplitudes and sine/cosine functions, males and females are similar; precisely, b) in Az1, there is a high significance, as running speed increases ($p<0.001$); however, c) in all amplitudes, the interaction between gender and speed is not significant. Furthermore, d) in all sine and cosine functions, no significances are found both as a function of running speed and in the interaction between gender and speed.

VARIABLE	Az1	Az2	Az3	Az4
Gender	p=NS	p=NS	p=NS	p=NS
Speed	<0.0001 **(p<0.001)**	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinz2	Sinz3	Sinz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

VARIABLE	Cosz2	Cosz3	Cosz4
Gender	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS
Interaction Gender/Speed	p=NS	p=NS	p=NS

Table 7.6c. Amplitude coefficients and sine/cosine functions in the lateral running direction (subjects aged 56 to 65).

Appendix 7.2

THE EXCURSION OF THE BODY CENTRE OF MASS

1. INTRODUCTION

Being able to calculate the average excursion of the body centre of mass along each movement direction, makes it possible (and easy) to graphically visualize how this pattern could change in various testing conditions (for instance, in different genders, as a function of age …).

In fact, as widely demonstrated in literature, in most cases females seem to have a larger pelvis than males. This is probably due to different physiological lifespan functions (for instance, pregnancy). As a result, the larger pelvis in females could require (and determine) a corresponding greater (lateral) excursion of the body centre of mass.

Among the others, we want to verify this hypothesis investigating whether the excursion of BCOM (independently of speed) could change as gender (males *versus* females) and age (children *versus* young adults *versus* elderly adults) vary.

Therefore, the following paragraphs will shortly illustrate and discuss this argument.

2. THE BODY CENTRE OF MASS EXCURSION: MAXIMUM AND MINIMUM

In order to compare males to females at different ages, we have normalized amplitude coefficients by the parameter Ay0 obtained in file *.res format (see chapter 6, par. 2). Indeed, Ay0 constitutes the average vertical position of the BCOM in each testing condition. This procedure makes it possible a) to adapt and cancel anthropometric differences, and b) to reach a homogeneity of data useful in further comparisons. To be precise:

- firstly, at every speed (walking and running), the average harmonic amplitude (A) has been divided by the corresponding average Ay0 value in the following way:

$$\mathrm{NormalizedA} = \frac{\mathrm{A}}{\mathrm{Ay0}} \quad \text{[Eq. 7.2]}$$

- this new value (Normalized A) measures mm/m;
- secondly, this procedure has been applied in all testing conditions, independently of movement direction.

Furthermore, to specifically investigate whether differences could be found between males and females, we mathematically calculated the average excursion of the BCOM in each movement direction. To be more specific, we obtained such excursions by applying Equation [6.9] previously discussed in chapter 6 (par. 5.2.2).

Means of both amplitudes and phases were used. However, instead of A, we put into the formula the value of Normalized A (Equation [7.2]) (see also Equation [7.3] below):

$$x(t)=\frac{Ax1}{Ay0}\bullet\sin(t+\varphi 1)+\frac{Ax2}{Ay0}\bullet\sin(2t+\varphi 2)+\frac{Ax3}{Ay0}\bullet\sin(3t+\varphi 3)+\frac{Ax4}{Ay0}\bullet\sin(4t+\varphi 4)+\frac{Ax5}{Ay0}\bullet\sin(5t+\varphi 5)+\frac{Ax6}{Ay0}\bullet\sin(6t+\varphi 6)$$

[Eq. 7.3a]

$$y(t)=\frac{Ay1}{Ay0}\bullet\sin(t+\varphi 1)+\frac{Ay2}{Ay0}\bullet\sin(2t+\varphi 2)+\frac{Ay3}{Ay0}\bullet\sin(3t+\varphi 3)+\frac{Ay4}{Ay0}\bullet\sin(4t+\varphi 4)+\frac{Ay5}{Ay0}\bullet\sin(5t+\varphi 5)+\frac{Ay6}{Ay0}\bullet\sin(6t+\varphi 6)$$

[Eq. 7.3b]

$$z(t)=\frac{Az1}{Ay0}\bullet\sin(t+\varphi 1)+\frac{Az2}{Ay0}\bullet\sin(2t+\varphi 2)+\frac{Az3}{Ay0}\bullet\sin(3t+\varphi 3)+\frac{Az4}{Ay0}\bullet\sin(4t+\varphi 4)+\frac{Az5}{Ay0}\bullet\sin(5t+\varphi 5)+\frac{Az6}{Ay0}\bullet\sin(6t+\varphi 6)$$

[Eq. 7.3c]

Finally, **only the first four harmonics** were taken into account, according to what explained in chapter 7, par. 2.3. In this way, forward, vertical and lateral excursions of the BCOM were obtained.

Some graphical examples are illustrated in Figure 7.20. The same colours described in chapter 1 (par. 3.6.2) were used. To be precise, in these graphs:

- the time period *t* ranges from 0 to 6.28 (100/degree) (step 0.79);
- forward excursion of BCOM ranges from -15 to 15 mm/m;
- vertical excursion of BCOM ranges from -60 to 60 mm/m;
- lateral excursion of BCOM ranges from -20 to 20 mm/m.

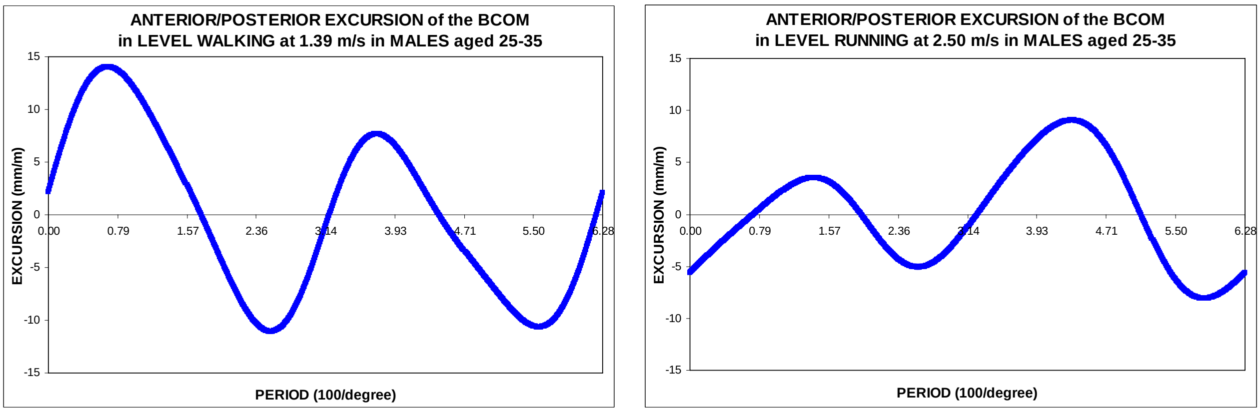

Figure 7.20a. Average anterior/posterior excursion of the BCOM during level walking at 1.39 m/s (on the left) and level running at 2.50 m/s (on the right), males aged 25 to 35.

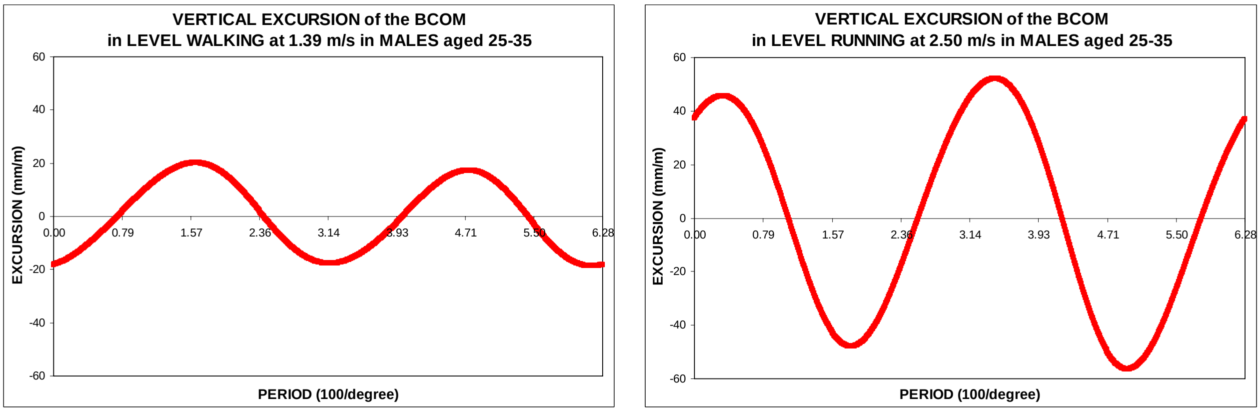

Figure 7.20b. Average vertical excursion of the BCOM during level walking at 1.39 m/s (on the left) and level running at 2.50 m/s (on the right), males aged 25 to 35.

Particularly, as already described in chapter 1 (par. 3.6.2), harmonic coefficients show that, both in forward (Figure 7.20a) and vertical (Figure 7.20b) directions, BCOM carries out periodical movements with the duration of a double step.

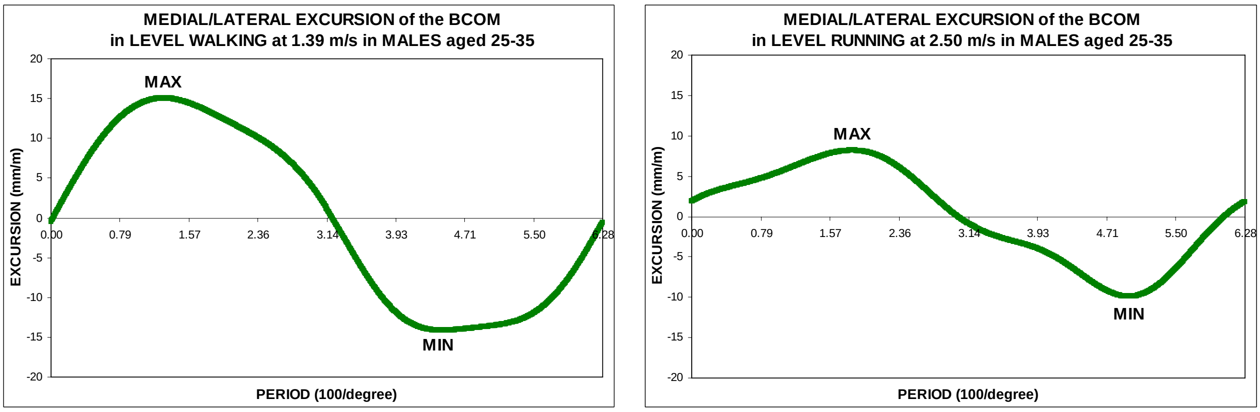

Figure 7.20c. Average medial/lateral excursion of the BCOM during level walking at 1.39 m/s (on the left) and level running at 2.50 m/s (on the right), males aged 25 to 35.

However, in lateral (Figure 7.20c) direction, BCOM carries out periodical movements with the duration of a single step.

In the last step of such an analysis, we have calculated both Maximum (MAX) and minimum (min) of all these excursions (an example has been illustrated in Figure 7.20c). By subtracting minimum from Maximum (Equation [7.4]), average range (or delta) of the BCOM excursion has been obtained, in each movement direction:

$$\mathrm{AverageRangeoftheBCOMExcursion} = \mathrm{Maximum} - \mathrm{minimum} \quad [\mathrm{Eq.}\ 7.4]$$

3. AVERAGE RANGE OF THE BODY CENTRE OF MASS EXCURSION

3.1. Introduction

By using Equation [7.4], average ranges of the BCOM excursion have been calculated in each movement direction (forward, vertical and lateral), in level gaits (walking and running). However, we decided not to calculate such parameters at gradient gaits because our main aim was to compare BCOM excursions as a function of both age and gender. In fact, our hypothesis was that differences between males and females aged 25 to 35 observed during level locomotion could be satisfactorily applied to different slopes. Moreover, our analysis makes it possible to investigate the pattern of these excursions independently of the factor speed. In all graphs below, results will be presented as mean ± standard deviation (S.D.). The patterns of excursion (independently of age) is represented in blue in males and in pink in females. The chosen independent variables were age group (y) and gender. The dependent variables were the BCOM excursions in each movement direction. Significances have been highlighted by the asterisks. Effect of gender on the dependent variable was assessed by using independent *t*-test (with Bonferroni correction). Moreover, effect of age was assessed by using a one-way ANOVA for unrelated measures. In addition, a post-*hoc* Bonferroni test was used to detect the strength of the associations between each dependent variable and age. SPSS software (version 12.0 for Windows) was used for statistical analysis (Zakeri et al., 2006; Houdijk et al., 2009). Average values of BCOM excursion (males and females in all age group, in both gaits) are contained in the enclosed CD (First Study, Appendix 7.1, Template level gait and Template gradient gait: files are catalogued as a function of age).

3.2. Body centre of mass excursion in forward direction

In forward direction, average range of the BCOM excursion ranges from 0 to 40 mm/m (walking and running), as shown in Figure 7.21:

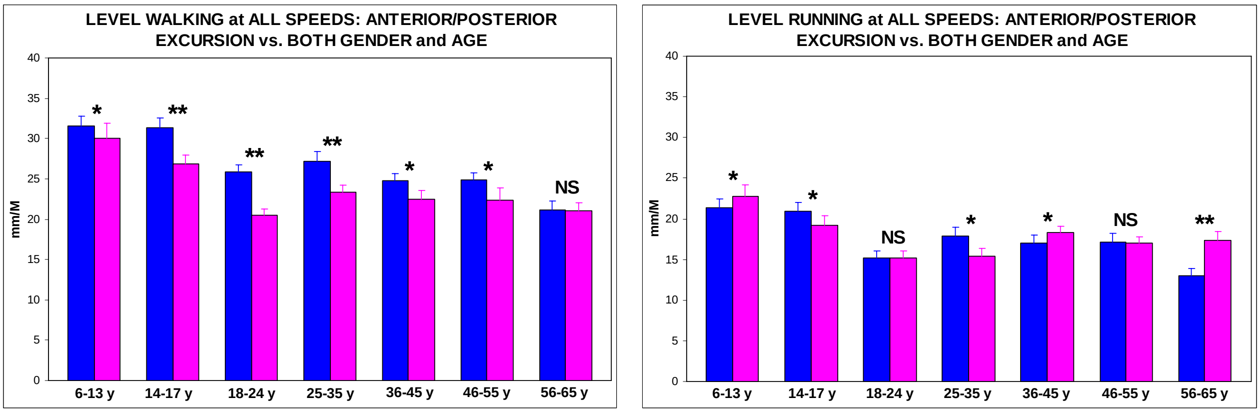

Figure 7.21. Average range of anterior/posterior excursion of the BCOM, males and females of all age groups (level walking, on the left; and level running, on the right).

A. Specifically, in **level walking** (left graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) males and females are wholly similar in subjects aged 56 to 65 (21.162 ± 1.031 mm/m - average values in males - *versus* 21.053 ± 0.972 mm/m - average value in females -); b) only slightly differences ($p<0.05$) are found in subjects aged 6 to 13 (31.584 ± 1.156 mm/m *versus* 30.003 ± 1.880 mm/m); 36 to 45 (24.755 ± 0.930 mm/m *versus* 22.498 ± 1.035 mm/m); and 46 to 55 (24.854 ± 0.940 mm/m *versus* 22.378 ± 1.523 mm/m); and c) the most significant differences ($p<0.01$) are found in 14 to 17 (31.302 ± 1.283 mm/m *versus* 26.850 ± 1.142 mm/m); 18 to 24 (25.884 ± 0.805 mm/m *versus* 20.545 ± 0.711 mm/m); and 25 to 35 (27.204 ± 1.143 mm/m *versus* 23.320 ± 0.935 mm/m) age groups;
- as a function of age: a) the highest BCOM excursion are in young subjects aged 6 to 17: 31.443 ± 1.219 mm/m (average value), in males; and 28.426 ± 1.511 mm/m, in females (independently of age); and b) among the other groups (from 18 to 65 years), there are no significant differences (24.772 ± 0.970 mm/m, in males; and 21.959 ± 1.035 mm/m, in females);
- finally, average BCOM excursion is slightly greater in males than in females ($p<0.05$), independently of age groups (26.678 ± 1.041 mm/m *versus* 23.807 ± 1.171 mm/m).

B. In **level running** (right graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) males and females are wholly similar in subjects aged 18 to 24 (15.147 ± 0.855 mm/m *versus* 15.195 ± 0.896 mm/m); and 46 to 55 (17.144 ± 1.071 mm/m *versus* 16.984 ± 0.748 mm/m); b) only slightly differences ($p<0.05$) are found in subjects aged 6 to 13 (21.391 ± 1.097 mm/m *versus* 22.797 ± 1.346 mm/m); 14 to 17 (20.953 ± 1.043 mm/m *versus* 19.152 ± 1.197 mm/m); 25 to 35 (17.845 ± 1.156 mm/m *versus* 15.422 ± 0.926 mm/m); and 36 to 45 (17.039 ± 0.974 mm/m *versus* 18.297 ± 0.796 mm/m); and c) the most significant differences ($p<0.01$) are found in 56 to 65 age group (13.028 ± 0.817 mm/m *versus* 17.296 ± 1.143 mm/m);
- as a function of age: no significance has been found among single age groups (17.507 ± 1.002 mm/m, in males; and 17.877 ± 1.007 mm/m, in females);
- finally, average BCOM excursion is slightly greater in males than in females ($p<0.05$), independently of age groups (100.533 ± 2.291 mm/m *versus* 95.597 ± 1.743 mm/m).

3.3. Body centre of mass excursion in vertical direction

In vertical direction, average range of the BCOM excursion ranges from 0 to 140 mm/m (walking and running), as shown in Figure 7.22:

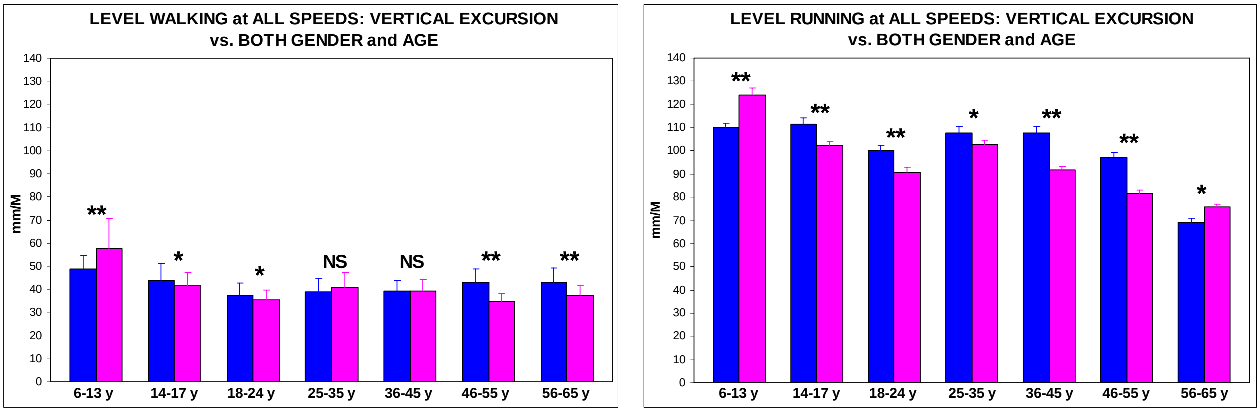

Figure 7.22. Average range of vertical excursion of the BCOM,
males and females of all age groups (level walking, on the left; and level running, on the right).

A. Specifically, in **level walking** (left graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) males and females are wholly similar in subjects aged 25 to 35 (39.096 ± 5.480 mm/m *versus* 40.681 ± 6.437 mm/m); and 36 to 45 (39.221 ± 4.790 mm/m *versus* 39.461 ± 4.766 mm/m); b) only slightly differences ($p<0.05$) are found in subjects aged 14 to 17 (44.007 ± 7.199 mm/m *versus* 41.558 ± 5.797 mm/m); and 18 to 24 (37.369 ± 5.271 mm/m *versus* 35.427 ± 4.210 mm/m); and c) the most significant differences ($p<0.01$) are in 6 to 13 (48.689 ± 6.012 mm/m *versus* 57.789 ± 12.827 mm/m); 46 to 55 (43.222 ± 5.590 mm/m *versus* 34.635 ± 3.628 mm/m); and 56 to 65 (43.071 ± 6.215 mm/m *versus* 37.519 ± 3.992 mm/m) age groups;
- as a function of age: no significance has been found among single age groups (42.096 ± 5.794 mm/m, in males; and 41.010 ± 5.951 mm/m, in females);
- finally, average BCOM excursion is very similar in both males and females, independently of age groups (42.096 ± 5.794 mm/m *versus* 41.010 ± 5.951 mm/m).

B. In **level running** (right graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) significant differences are found in all age groups: subjects aged 6 to 13 (110.155 ± 1.695 mm/m *versus* 124.032 ± 2.982 mm/m), $p<0.001$; 14 to 17 (111.513 ± 2.591 mm/m *versus* 102.466 ± 1.513 mm/m), $p<0.001$; 18 to 24 (100.296 ± 2.307 mm/m *versus* 90.658 ± 2.405 mm/m), p<0.001; 25 to 35 (107.620 ± 2.878 mm/m *versus* 102.758 ± 1.572 mm/m), p<0.01; 36 to 45 (107.909 ± 2.632 mm/m *versus* 91.863 ± 1.287 mm/m), p<0.001; 46 to 55 (97.047 ± 2.348 mm/m *versus* 81.592 ± 1.342 mm/m), p<0.001; and 56 to 65 (69.194 ± 1.589 mm/m *versus* 75.809 ± 1.104 mm/m), p<0.01. The older is the subject, the shorter is the corresponding BCOM excursion;

- as a function of age: only slightly differences (p<0.05) have been found among single age groups (100.533 ± 2.291 mm/m, in males; and 95.957 ± 1.743 mm/m, in females).

3.4. Body centre of mass excursion in lateral direction

In lateral direction, average range of the BCOM excursion ranges from 0 to 45 mm/m (walking and running), as shown in Figure 7.23:

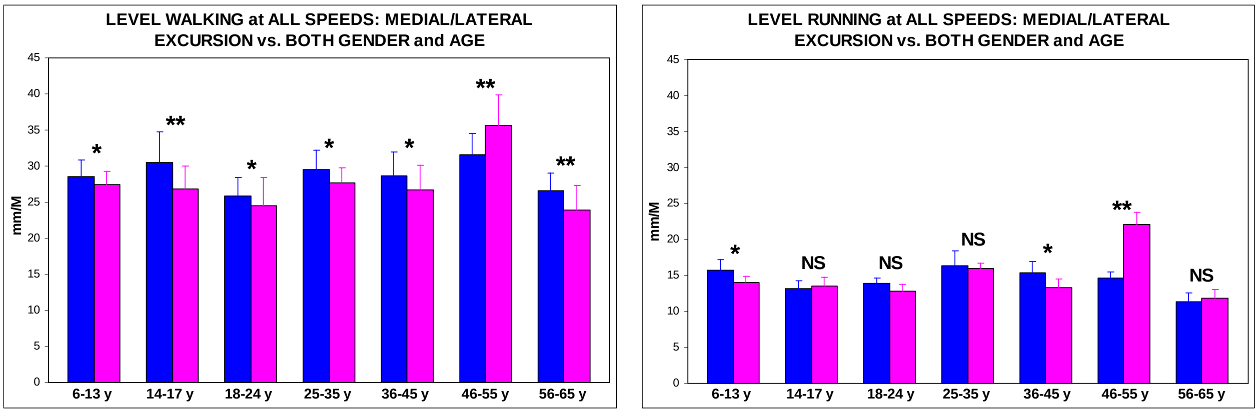

Figure 7.23. Average range of medial/lateral excursion of the BCOM, males and females of all age groups (level walking, on the left; and level running, on the right).

A. Specifically, in **level walking** (left graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) only slightly differences (p<0.05) are found in subjects aged 6 to 13 (28.529 ± 2.362 mm/m *versus* 27.471 ± 1.783 mm/m); 18 to 24 (25.794 ± 2.652 mm/m *versus* 24.500 ± 3.867 mm/m); 25 to 35 (29.537 ± 2.604 mm/m *versus* 27.682 ± 2.093 mm/m); and 36 to 45 (28.601 ± 3.379 mm/m *versus* 26.668 ± 3.442 mm/m); and b) the most significant differences (p<0.01) are in 14 to 17 (30.503 ± 4.231 mm/m *versus* 26.795 ± 3.158 mm/m); 46 to 55 (31.593 ± 2.934 mm/m *versus* 35.638 ± 4.211 mm/m); and 56 to 65 (27.909 ± 2.354 mm/m *versus* 23.952 ± 3.389 mm/m) age groups;
- as a function of age: no significance has been found among single age groups (26.628 ± 2.931, in males; and 27.529 ± 3.135, in females).

B. Specifically, in **level running** (right graph), our results show that there are slight changes as a function of both gender and age, independently of speed. In detail:

- as a function of gender: a) males and females are wholly similar in subjects aged 14 to 17 (13.170 ± 1.071 mm/m *versus* 13.590 ± 1.155mm/m); 18 to 24 (13.921 ± 0.684 mm/m *versus* 12.774 ± 1.041 mm/m); 25 to 35 (16.390 ± 1.990 mm/m *versus* 15.960 ± 0.787 mm/m); and 56 to 65 (11.310 ± 1.225 mm/m *versus* 11.796 ± 1.219 mm/m); b) only slightly differences ($p<0.05$) are found in subjects aged 6 to 13 (15.678 ± 1.541 mm/m *versus* 14.082 ± 0.848 mm/m); and 36 to 45 (15.356 ± 1.547 mm/m *versus* 13.256 ± 1.214 mm/m); and c) the most significant differences ($p<0.01$) are in 46 to 55 age group (14.596 ± 0.869 mm/m *versus* 22.051 ± 1.768 mm/m);
- as a function of age: no significance has been found among single age groups (14.346 ± 1.274, in males; and 14.787 ± 1.148, in females).

4. DISCUSSION

The mathematical analysis of BCOM excursion (in all movement directions) partially concurs with our initial hypothesis. Indeed, only in some isolated cases females seem to have a greater excursion compared to males:

a) in the forward direction: level running in females aged 56 to 65;
b) in the vertical direction: level walking and running in females aged 6 to 13;
c) in the lateral direction: level walking and running in females aged 46 to 55.

This higher excursion could be probably related to females having a larger pelvis.

However, it seems a bit strange that such a significance has been found only in few testing conditions. We suggest that it could depend on similar anthropometric dimensions between males and females.

Moreover, it is important to underline the greatest vertical BCOM excursion, especially in running, compared to the other movement directions. This fundamental result wholly satisfies our hypothesis that BCOM raises and lifts as both walking and running speed increases (see also chapter 7, par. 3 onwards).

Finally, the contribution of the harmonic coefficients of both the fifth and the sixth order will not evidently modify average BCOM excursions since their relative importance.

In order to complete our analysis, we have to remember that two other ways to investigate the BCOM excursion could be applied:

1. when normalizing data, instead of the average Ay0 (see par. 7.2 above), it could be possible to use the average height. Specifically, this height will be derived from anthropometric

dimensions in each age group. However, in our analysis, we decided to apply the first method in order to emphasize the results obtained by using the mathematical method based upon Fourier Series. Other than that, the application of both the two methods will reach similar results;

2. the BCOM excursion could be calculated by using the mathematical derivation of the second order, as well. This will lead to same results and conclusions.

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Zakeri I., Puyau M.R., Adolph A.L., Vohra F.A., Butte N.F. (2006) Normalization of energy expenditure data for differences in body mass or composition in children and adolescents. *Journal of Nutrition* 136: 1371-1376.

Chapter 8

SYMMETRY INDEX

1. MATHEMATICAL DEFINITION

1.1. Introduction

Symmetry plays a key role in simplifying the control of humans, animals and legged robots and in giving them the ability to run and balance (Raibert, 1986; Griffin et al., 1995).

As stated in literature, traditional parameters used to assess gait asymmetries, e.g. a) laterality (Sadeghi et al., 2000), b) joint range of motion or c) freedom degrees (Shorter et al., 2008), fall to provide insight regarding timing and magnitude of movement deviations during the gait cycle.

So that a new approach for quantifying aspects of gait asymmetry is necessary a) to compare kinematic symmetry across various experimental groups (Karamanidis et al., 2003; Herbin et al., 2004) and b) to investigate both normal and clinical settings (Shorter et al., 2008; Brouwer et al., 2009; Starke et al., 2009).

1.2. Mathematical equation

Therefore, Symmetry Index we calculated could help in solving these problems (Robinson et al., 1987; Giakas et al., 1997; Draper, 2000). To be precise, Symmetry Index (SI) is a mathematical index, namely representing the spatial differences, in BCOM trajectory, between the two strides.

Indeed, it contains and summarizes the most important information regarding symmetry/asymmetry in each movement direction. It can be also used to compare the behaviour of single harmonic amplitude coefficients (see Equations [8.1], [8.2] and [8.3]). However, it does not give information about sides (i.e. right or left) or direction (i.e. forward or backward) of locomotion.

As previously described (see also chapter 6), harmonic coefficients were obtained from the average values appreciated for any testing condition (gender, age, type of locomotion, speed and gradient). Particularly, we have expressed the Symmetry Index as:

$$\mathrm{SIx} = \frac{\mathrm{Ax2} + \mathrm{Ax4} + \mathrm{Ax6}}{\Sigma \mathrm{Ax_i}} \quad [\text{Eq. } 8.1]$$

$$\mathrm{SIy} = \frac{\mathrm{Ay2} + \mathrm{Ay4} + \mathrm{Ay6}}{\Sigma \mathrm{Ay_i}} \quad [\text{Eq. } 8.2]$$

$$\mathrm{SIz} = \frac{\mathrm{Az1} + \mathrm{Az3} + \mathrm{Az5}}{\Sigma \mathrm{Az_i}} \quad [\text{Eq. } 8.3]$$

where *SIx*, *SIy* and *SIz* are the Symmetry Index in the anterior/posterior, vertical and medial/lateral direction, respectively; *A2*, *A4* and *A6* are the symmetrical harmonic coefficients in the anterior/posterior and vertical directions (see also chapter 6, par. 5.2.3); *A1*, *A3* and *A5* are the symmetrical harmonic coefficients in the medial/lateral direction (see also chapter 6, par. 5.2.3); and $\sum A_i$ is the mathematical sum of each harmonic coefficient.

As a result, we have appreciated Symmetry Index for each movement direction both in walking and running (level and gradient condition). In this way, it was possible to compare the values of this a-dimensional variable in different testing condition as age, speed and gradient changes.

Single values of Symmetry Index (males and females in all age group, in both gaits) are contained in the enclosed CD (First Study, Chapter 8, Symmetry Index in level and gradient gaits).

2. STATISTICAL ANALYSIS

Statistical analysis was performed by using each subject Symmetry Index value.

Results will be presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The chosen independent variables were age group (y), progression speed (m/s) and gradient (%). The dependent variable was the Symmetry Index (SI).

Effects of gender and age on the dependent variable were assessed by using a one-way ANOVA for unrelated measures. In addition, a post-*hoc* Bonferroni test was used to detect the strength of the associations between each dependent variable and gender/age.

Moreover, effects of speed and gradient were assessed by using a one-way ANOVA for related measures. In addition, a post-*hoc* paired *t*-test (with Bonferroni correction) was used to detect differences between each dependent variable and speed/gradient.

SPSS software (version 12.0 for Windows) was used for statistical analysis (Zakeri et al., 2006; Houdijk et al., 2009). Specific results of this statistical analysis are presented and described in the Appendix 8.1. Finally, specifically, single values of Symmetry Index are contained in the enclosed CD (First Study, Chapter 8, Statistical analysis).

3. GRAPH LEGEND

In each graph, points represent mean values obtained by grouping the same age subjects in the different testing condition. Lines represent the simple graphic amalgamation of all data. Vertical bars represent the standard deviation of the minor and major average values (mean ± S.D.), respectively. To be more specific, graphs were drawn up as follows:

a) As a function of age:

- age group (y) is the independent variable. For each group, we have plotted the data corresponding to the median (middle of the distribution) of age (i.e. the middle of 6 to 13 age group is 9 y; the middle of 14 to 17 age group is 16 y and so on);
- values of each dependent variable in level/gradient walking are plotted in blue at the speed of 0.83 m/s; in yellow at the speed of 1.11 m/s; in dark green at the speed of 1.39 m/s; in ski-blue at the speed of 1.67 m/s and in red at the speed of 1.94 m/s;
- values of each dependent variable in level/gradient running are plotted in green at the speed of 1.94 m/s; in orange at the speed of 2.22 m/s; in white at the speed of 2.50 m/s; in pink at the speed of 2.78 m/s and in blue at the speed of 3.06 m/s.

b) As a function of speed:

- walking speed (m/s) is the independent variable. The average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 (see also chapter 6, par. 2.1) have been considered;
- for each age group there is a corresponding colour (Table 8.1).

SUBJECTS: AGE GROUPS	CORRESPONDING COLOUR
6 - 13 y	BLACK
14 - 17 y	RED
18 - 24 y	DARK GREEN
25 - 35 y	ORANGE
36 - 45 y	SKI-BLUE
46 - 55 y	BLU
56 - 65 y	PINK

Table 8.1. Colour *per* each age group.

c) As a function of gradient:

- gradient (%) is the independent variable;
- values of each dependent variable in gradient walking and running are the same presented in point a).

4. RESULTS OF OUR EXPERIMENTS

4.1. Introduction

In theory, Symmetry Index could range from 0 (= no symmetry) to 1 (= complete symmetry).

However, in our experiments, in each movement direction, SI ranges from a minor value of 0.4 (= no symmetry) to a major one of 1 (= complete symmetry; step 0.1). Consequently, we have decided to put 0.4 as the minor value corresponding to the absence of symmetry. To be more specific, three different situations are possible:

1. if there is no symmetry, then Symmetry Index equals 0.4;
2. if there is an increasing asymmetry, then Symmetry Index ranges from 0.4 to 1;
3. if there is a complete symmetry, then Symmetry Index equals 1.

4.2. Symmetry Index as a function of age in level walking

A. In the **anterior/posterior** direction (Figure 8.1), our results show that:

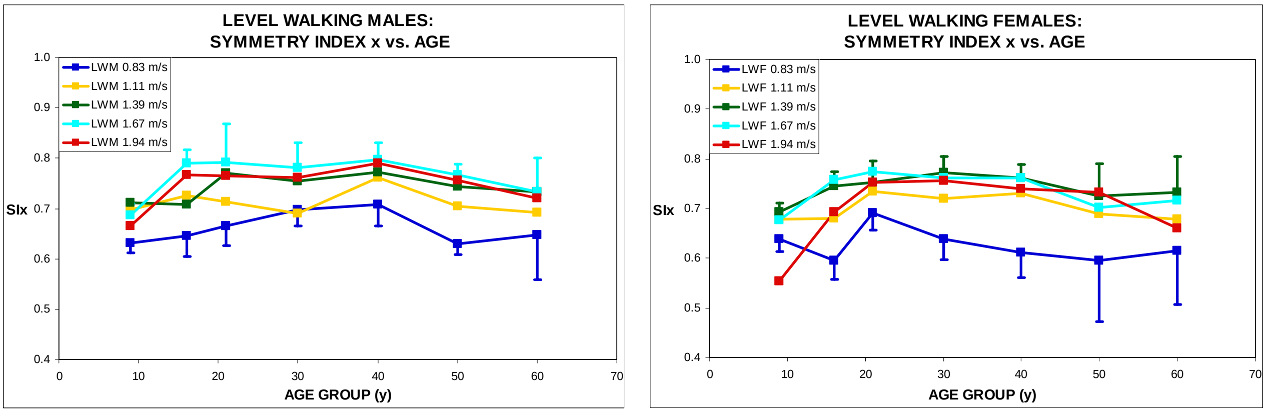

Figure 8.1. Symmetry Index in the anterior/posterior direction as a function of age in level walking, males (on the left) and females (on the right).

- both in males and females, Symmetry Index x is slightly lower in young children aged 6 to 13 (0.678 ± 0.046, independently of gender, age and speed, $p<0.05$);
- in males (left graph), this pattern is evident if age group 6 to 13 (0.668 ± 0.046, at all the investigated speeds) is compared to age group 14 to 45 (0.743 ± 0.045);
- in females (right graph), it becomes more clear if age group 6 to 13 (0.648 ± 0.039) is compared to age group 14 to 35 (0.722 ± 0.040);
- these patterns occur similarly at each speed;
- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index. As a consequence, there is an increasing symmetry with speed;
- on average, its values in males (0.725 ± 0.047, independently of age and speed) are similar to female values (0.700 ± 0.051);
- in general, in all age group, there is no complete symmetry. This pattern occurs similarly at each speed.

In conclusion, we could state that, both in males and females, this represents the most asymmetrical movement direction.

Specific results of the statistical analysis (with relevance) are shown in par. 2.1, Table 8.2a (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.2), our results show that:

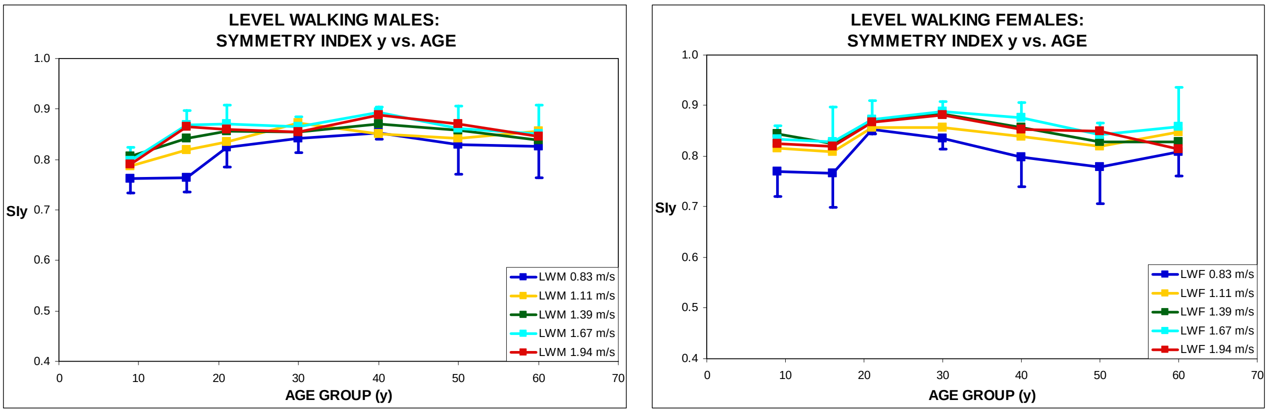

Figure 8.2. Symmetry Index in the vertical direction as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), Symmetry Index y is lower in young children aged 6 to 13 (0.788 ± 0.044, at all the investigated speeds, $p<0.001$). In the other age groups, there is substantially a similar pattern (0.842 ± 0.036, independently of age and speed);
- in females (right graph), it is slightly greater in young adults aged 18 to 25 (0.864 ± 0.031; at all the investigated speeds, $p<0.05$). Moreover, it little decreases with age ($p<0.05$);
- on average, its values in males are similar to female values (0.837 ± 0.040);
- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index. As a consequence, there is an increasing symmetry with speed.

To sum up, along this direction, in all age groups, there is a much more symmetry.

Specific results of the statistical analysis (with relevance) are shown in par. 2.2, Table 8.2b (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.3), our results show that:

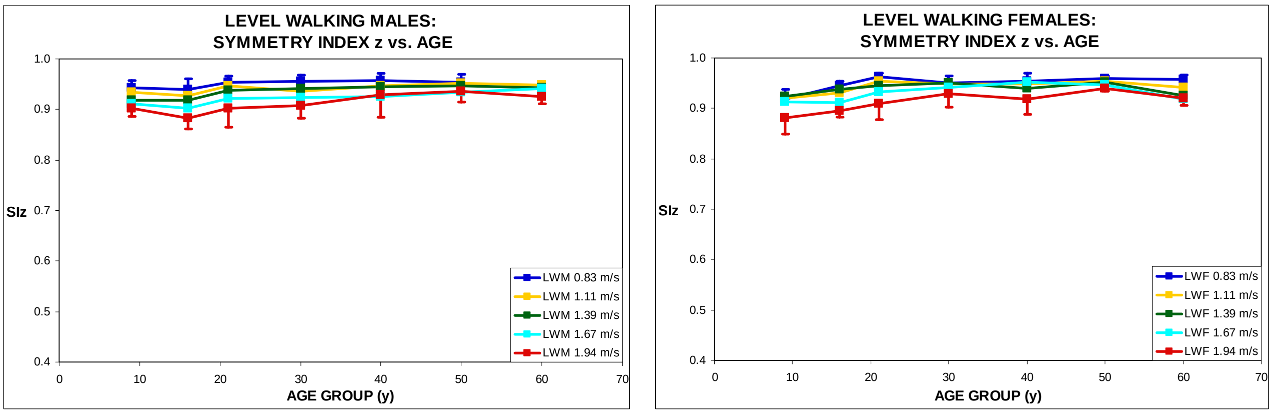

Figure 8.3. Symmetry Index in the medial/lateral direction as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), Symmetry Index z is slightly higher in subjects aged 6 to 45 (0.931 ± 0.021, at all the investigated speeds, $p<0.05$);
- in females (right graph), it is slightly lower in subjects aged 6 to 17 (0.917 ± 0.018, $p<0.05$);
- these patterns are quite similar at each speed;
- on average, its values in males (0.933 ± 0.021, independently of age and speed) are similar to female values (0.935 ± 0.017);
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index. As a consequence, there is a decreasing symmetry with speed;
- in general, along this direction, in all age group, there is its highest value.

In conclusion, we could state that, both in males and females, this is the most symmetrical movement direction.

Specific results of the statistical analysis (with relevance) are shown in par. 2.3, Table 8.2c (see Appendix 8.1).

4.3. Symmetry Index as a function of age in level running

A. In the **anterior/posterior** direction (Figure 8.4), our results show that:

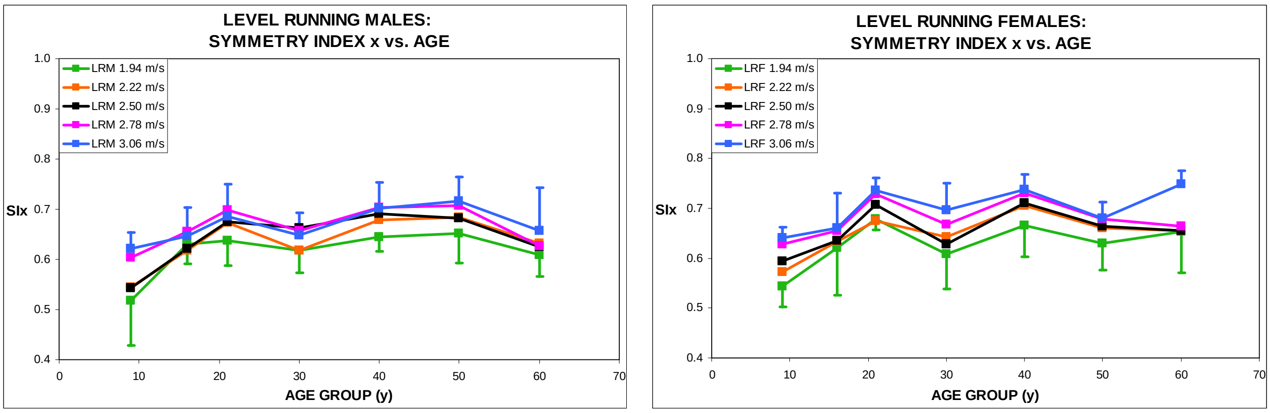

Figure 8.4. Symmetry Index in the anterior/posterior direction as a function of age in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index x is lower in subjects aged 6 to 17 (0.600 ± 0.050, at all the investigated speeds, in males; and 0.618 ± 0.049, in females, $p<0.001$);
- furthermore, in males, it is lower in elderly subjects aged 56 to 65 (0.630 ± 0.073, at all the investigated speeds; $p<0.01$);
- these patterns are quite similar at each speed;

- on average, Symmetry Index is slightly higher in females (0.663 ± 0.041, independently of age and speed) than in males (0.645 ± 0.050);
- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index. As a consequence, there is an increasing symmetry with speed;
- in general, in all age group, there is no complete symmetry. This pattern occurs similarly at each speed.

In conclusion, we could state that, both in males and females, this represents the most asymmetrical movement direction.

Specific results of the statistical analysis (with relevance) are shown in par. 3.1, Table 8.3a (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.5), our results show that:

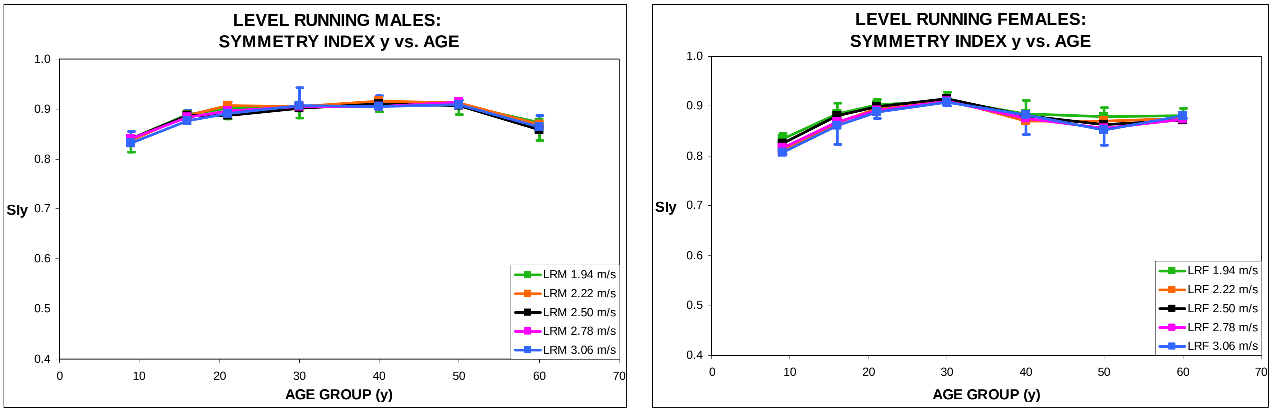

Figure 8.5. Symmetry Index in the vertical direction as a function of age in level running, males (on the left) and females (on the right).

- in males (left graph), Symmetry Index y is lower both in subjects aged 6 to 17 (0.862 ± 0.017, at all the investigated speeds, $p<0.001$) and in elderly adults aged 56 to 65 (0.865 ± 0.034, $p<0.001$). However, it does not significantly change in the other age groups (0.905 ± 0.020);
- in females (right graph), it is lower both in children and young adults aged 6 to 24 (0.862 ± 0.019, $p<0.001$) and in adults aged 36 to 65 (0.873 ± 0.026, $p<0.001$);
- these pattern are quite similar at each speed;
- on average, its values in males (0.887 ± 0.021, independently of age and speed) are similar to female values (0.874 ± 0.021). However, in females it is lower in adults aged 36 to 55;
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index. As a consequence, there is a little decreasing symmetry with speed.

To sum up, along this direction, in all age group, there is a much more symmetry.

Specific results of the statistical analysis (with relevance) are shown in par. 3.2, Table 8.3b (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.6), our results show that:

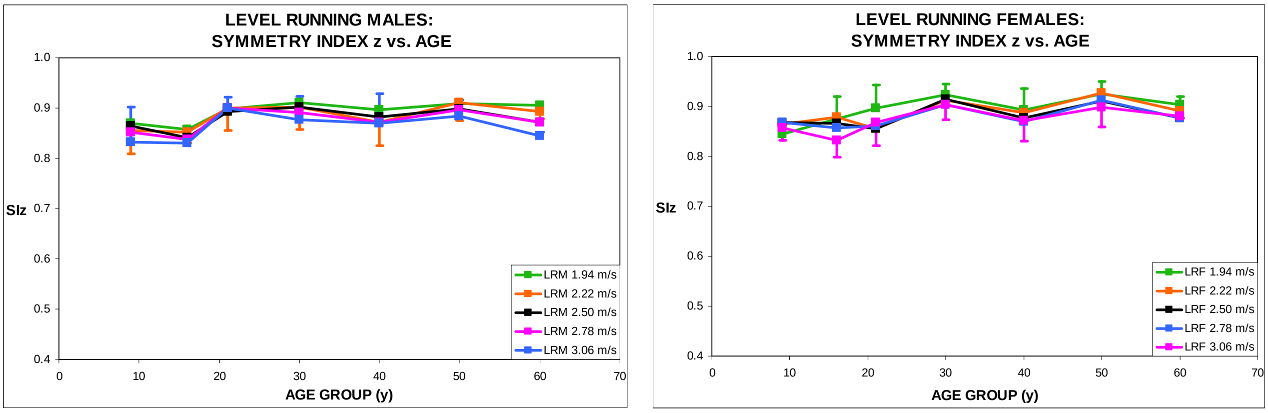

Figure 8.6. Symmetry Index in the medial/lateral direction as a function of age in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index z is lower in young subjects aged 6 to 17 (0.849 ± 0.040, at all the investigated speeds, in males; and 0.861 ± 0.026 in females, $p<0.001$);
- these patterns are quite similar at each speed;
- on average, its values are slightly lower in males (0.878 ± 0.037, independently of age and speed) than in females (0.883 ± 0.030);
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index. As a consequence, there is a little decreasing symmetry with speed;
- in general, along this direction, in all age group, there is its highest value.

In conclusion, as shown above in walking condition, this is the most symmetrical movement direction.

Specific results of the statistical analysis (with relevance) are shown in par. 3.3, Table 8.3c (see Appendix 8.1).

4.4. Symmetry Index as a function of speed in level walking

A. In the **anterior/posterior** direction (Figure 8.7), our results show that:

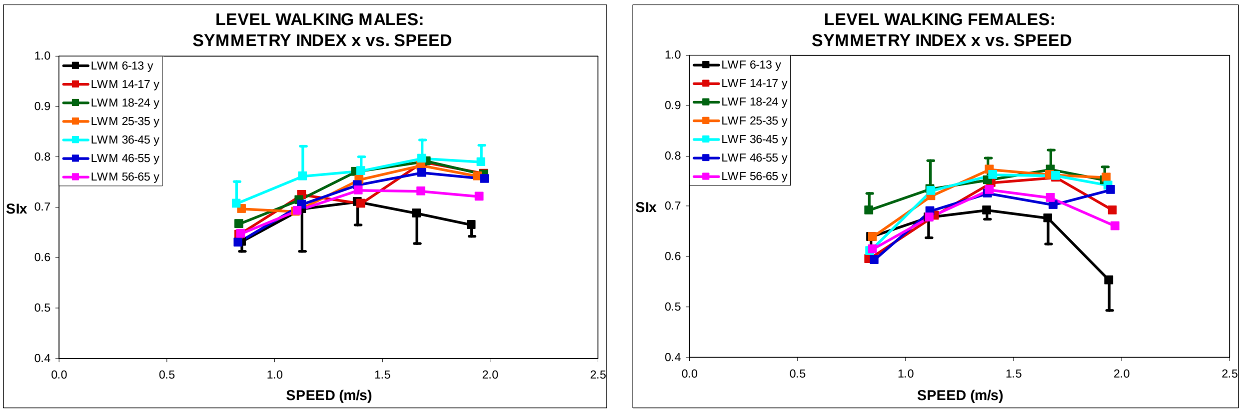

Figure 8.7. Symmetry Index in the anterior/posterior direction as a function of speed in level walking, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index x increases with speed only from 0.83 to 1.11 m/s (from 0.661 ± 0.041 to 0.712 ± 0.056, independently of age, in males; and from 0.626 ± 0.060 to 0.702 ± 0.053, in females, $p<0.001$) and from 1.11 to 1.39 m/s (from 0.712 ± 0.056 to 0.742 ± 0.054, in males, $p<0.001$; and from 0.702 ± 0.053 to 0.741 ± 0.040, in females, $p<0.01$);
- in males, from 1.39 to 1.94 m/s, it does not significantly change with speed (from 0.742 ± 0.054 to 0.747 ± 0.043). This pattern is quite similar in each age group;
- however, in females, the only exceptions are subjects aged 6 to 17 and elderly adults aged 56 to 65. In fact, it little decreases from 1.67 to 1.94 m/s (from 0.717 ± 0.041 to 0.622 ± 0.050, in females aged 6 to 17, $p<0.01$; and from 0.716 ± 0.069 to 0.660 ± 0.058, in females aged 56 to 65, $p<0.05$);
- on average, it is slightly lower in young children aged 6 to 13 (0.666 ± 0.023, in males; and 0.553 ± 0.060, in females) and in elderly adults aged 56 to 65 (0.721 ± 0.068, in males; and 0.660 ± 0.058, in females): this pattern is especially evident at the highest walking speed (= 1.94 m/s).

Specific results of the statistical analysis (with relevance) are shown in par. 4.1, Table 8.4a and 8.4b (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.8), our results show that:

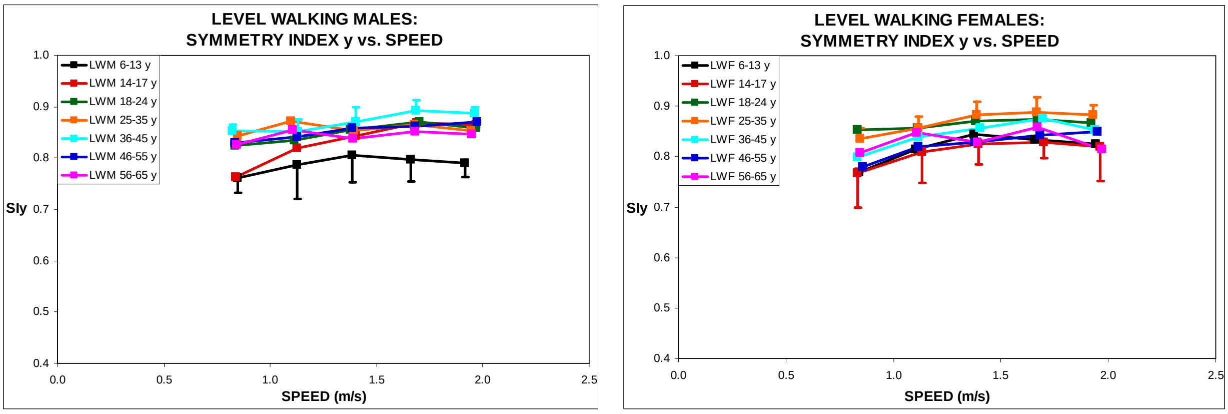

Figure 8.8. Symmetry Index in the vertical direction as a function of speed in level walking, males (on the left) and females (on the right).

- both in males and females, Symmetry Index y increases with speed only from 0.83 to 1.11 m/s (from 0.814 ± 0.037 to 0.837 ± 0.039, independently of age, in males, $p<0.001$; and from 0.802 ± 0.047 to 0.835 ± 0.035, in females, $p<0.001$);
- furthermore, in male and female young subjects aged 6 to 13 (from 0.803 ± 0.050 to 0.825 ± 0.051, in males; and from 0.812 ± 0.045 to 0.835 ± 0.033, in females) and in female adults aged 36 to 45 (from 0.839 ± 0.047 to 0.856 ± 0.044), it also increases from 1.11 to 1.39 m/s ($p<0.05$);
- however, above these speeds, it does not significantly change with speed;
- on average, in males (left graph), it is slightly lower in young children aged 6 to 13 (0.838 ± 0.019, at all the investigated speeds); this pattern is not so evident in females (right graph).

Specific results of this statistical analysis (with relevance) are shown in par. 4.2, Table 8.4c and 8.4d (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.9), our results show that:

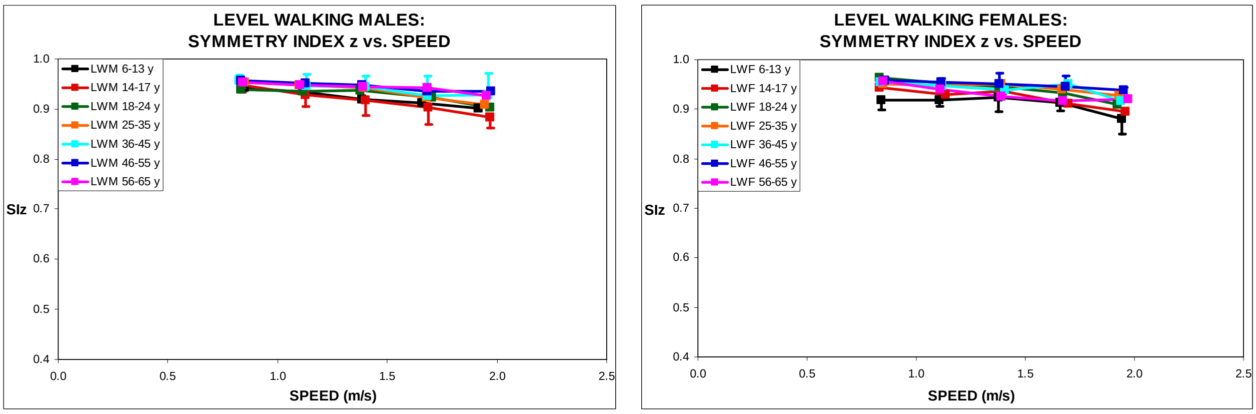

Figure 8.9. Symmetry Index in the medial/lateral direction as a function of speed in level walking, males (on the left) and females (on the right).

- in males (left graph), Symmetry Index z decreases with speed (from 0.950 ± 0.015 to 0.912 ± 0.025, independently of age and speed, p<0.001);
- in females (right graph), it does not significantly change with speed up to 1.39 m/s. Above this speed, it then decreases with speed (from 0.939 ± 0.019 to 0.913 ± 0.022, p<0.01);
- these patterns are similar in all age groups.

Specific results of this statistical analysis (with relevance) are shown in par. 4.3, Table 8.4e (see Appendix 8.1).

4.5. Symmetry Index as a function of speed in level running

A. In the **anterior/posterior** direction (Figure 8.10), our results show that:

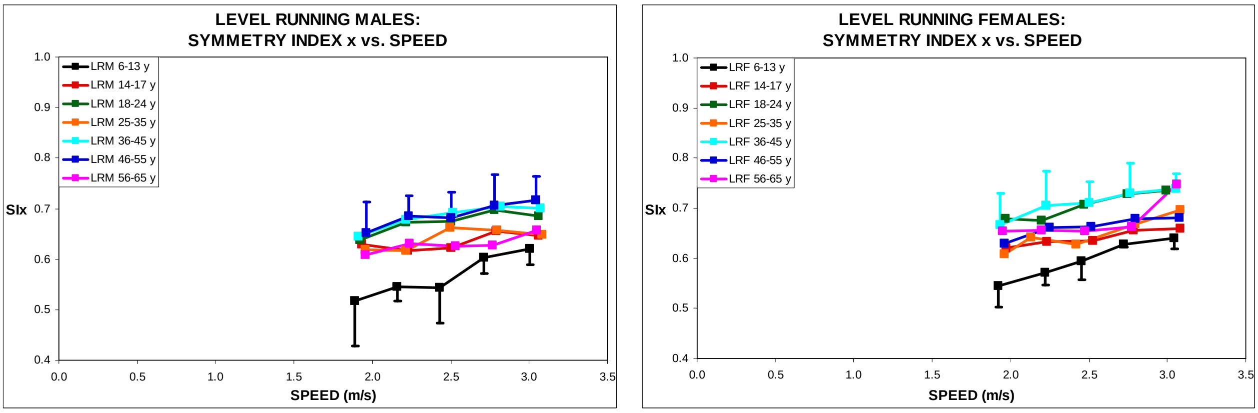

Figure 8.10. Symmetry Index in the anterior/posterior direction as a function of speed in level running, males (on the left) and females (on the right).

- in male and female young subjects aged 6 to 13, Symmetry Index x increases at all speeds (from 0.517 ± 0.089 to 0.621 ± 0.033, in males; and 0.544 ± 0.043 to 0.641 ± 0.022, in females, p<0.001);
- however, both in males and females aged 14 to 55, it little increases with speed up to 2.50 m/s (from 0.633 ± 0.041 to 0.679 ± 0.039; and from 0.643 ± 0.063 to 0.696 ± 0.055; p<0.01). Above this speed, it does not change with speed;
- furthermore, in male and female elderly adults aged 56 to 65, it does not significantly change with speed;
- on average, both in males (left graph) and females (right graph), it is lower in young children aged 6 to 13. Particularly, it is 0.566 ± 0.050 in males aged 6 to 13 and 0.659 ± 0.050 in males aged 14 to 65, at all the investigated speed; it is 0.595 ± 0.026 in females aged 6 to 13 and 0.674 ± 0.058 in females aged 14 to 65.

Specific results of this statistical analysis (with relevance) are shown in par. 5.1, Table 8.5a and 8.5b (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.11), our results show that:

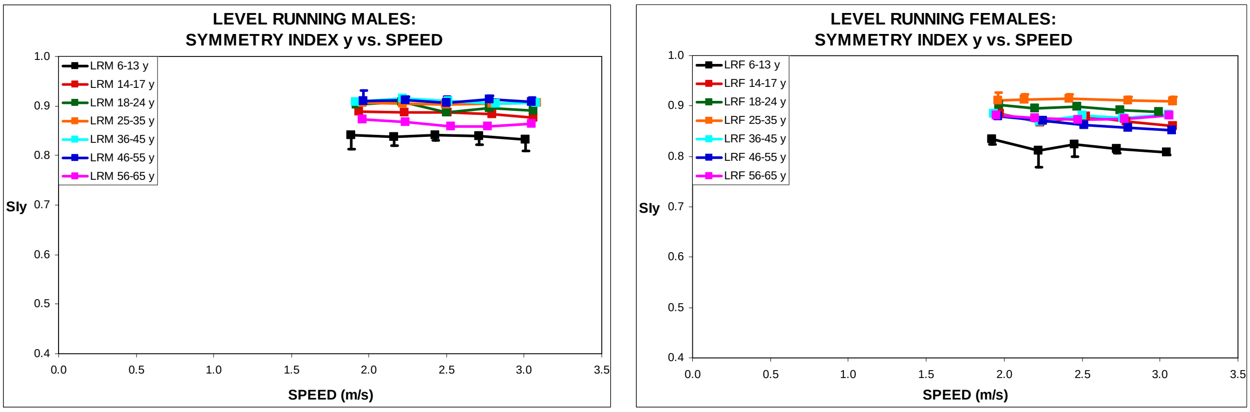

Figure 8.11. Symmetry Index in the vertical direction as a function of speed in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index y does not significantly change with speed. The only exception is in females aged 46 to 55. In this case, it little decreases with speed (from 0.879 ± 0.017 to 0.852 ± 0.030, $p<0.01$);
- on average, both in males and females, it is lower in young children (aged 6 to 13). Particularly, it is 0.838 ± 0.019 in males aged 6 to 13, and 0.895 ± 0.021 in males aged 14 to 65, at all the investigated speed; it is 0.819 ± 0.017 in females aged 6 to 13, and 0.883 ± 0.021 in females aged 14 to 65.

Specific results of this statistical analysis (with relevance) are shown in par. 5.2, Table 8.5c (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.12), our results show that:

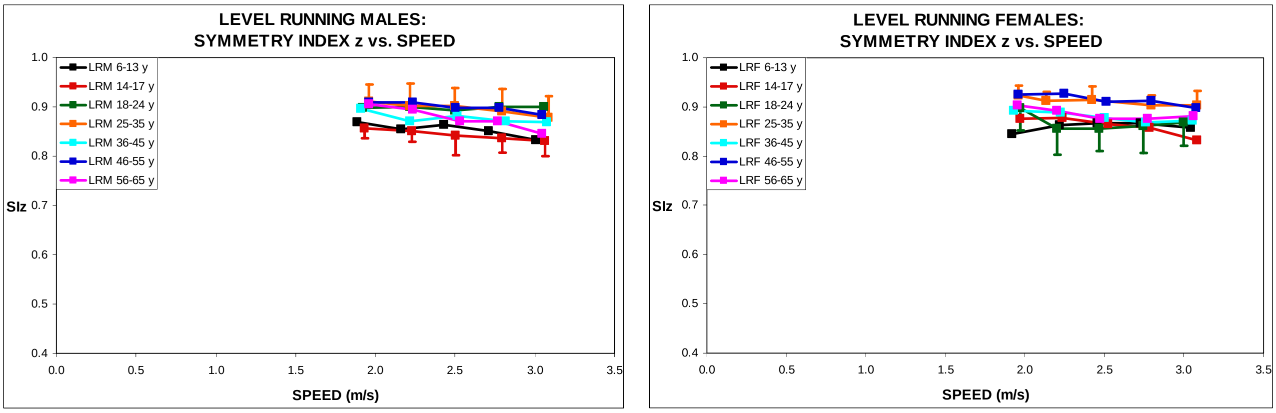

Figure 8.12. Symmetry Index in the medial/lateral direction as a function of speed in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index z little decreases with speed (from 0.892 ± 0.034 to 0.863 ± 0.038, in males, independently of age; and from 0.895 ± 0.030 to 0.873 ± 0.032, in females, $p<0.01$);
- these patterns are similar in each age group.

Specific results of this statistical analysis (with relevance) are shown in par. 5.3, Table 8.5d (see Appendix 8.1).

4.6. Symmetry Index as a function of gradient in walking

A. In the **anterior/posterior** direction (Figure 8.13), our results show that:

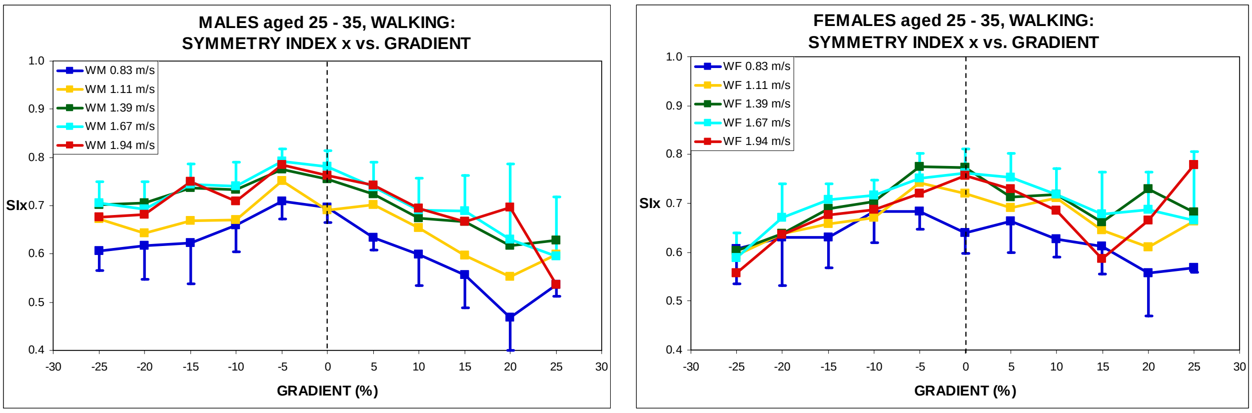

Figure 8.13. Symmetry Index in the anterior/posterior direction as a function of gradient in walking, males (on the left) and females (on the right).

- both in males and females, Symmetry Index x is dependent on gradient;
- in males (left graph), it little increases with gradient up to -5% (from 0.673 ± 0.044 to 0.763 ± 0.032, at all the investigated speeds); above this slope, it significantly decreases (from 0.763 ± 0.032 to 0.579 ± 0.063);
- however, in females (right graph), it little increases with gradient up to the level condition (from 0.590 ± 0.070 to 0.730 ± 0.044); above this slope, it significantly decreases (from 0.730 ± 0.044 to 0.671 ± 0.089);
- on average, in the downhill gait, in males (0.702 ± 0.054, at all the investigated speeds) it is higher than the corresponding female values (0.666 ± 0.065);
- on average, in both the level and the uphill gaits, its values in males (0.737 ± 0.044 and 0.636 ± 0.059, respectively) are similar to female values (0.730 ± 0.044 and 0.672 ± 0.067);
- Symmetry Index patterns are quite similar at each speed;
- gradient walking at the lowest speed (0.83 m/s) and at the highest one (1.94 m/s) are the most variable and asymmetrical conditions;

- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index;
- in general, along this direction there is no complete symmetry.

In conclusion, both in males and females, this represents the most asymmetrical movement direction.

Specific results of this statistical analysis (with relevance) are shown in par. 6.1, Table 8.6a (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.14), our results show that:

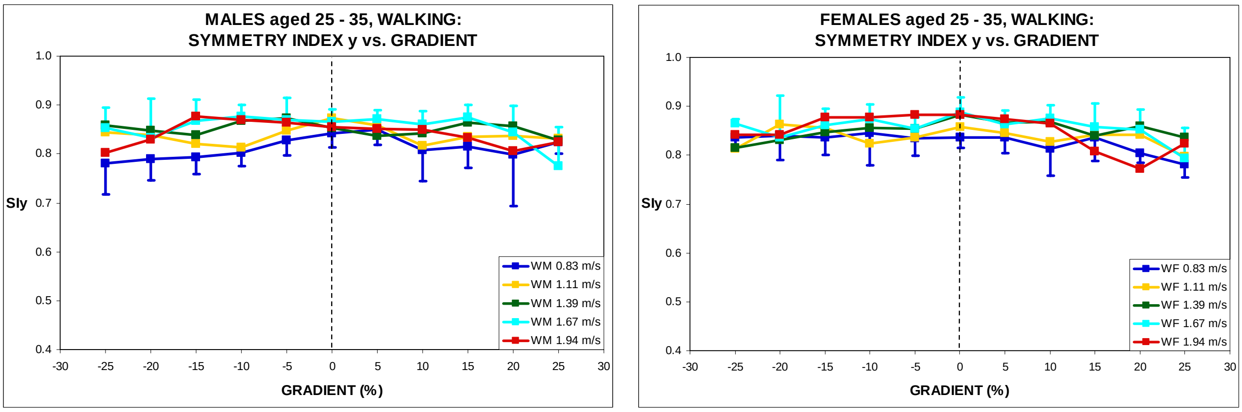

Figure 8.14. Symmetry Index in the vertical direction as a function of gradient in walking, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index y is only little dependent on gradient;
- Symmetry Index patterns are quite similar at each speed;
- on average, its values in males (0.810 ± 0.042, independently of speed and gradient) are slightly lower than corresponding female values (0.844 ± 0.039);
- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index.

In conclusion, along this direction there is a much more symmetry.

Specific results of this statistical analysis (with relevance) are shown in par. 6.2, Table 8.6b (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.15), our results show that:

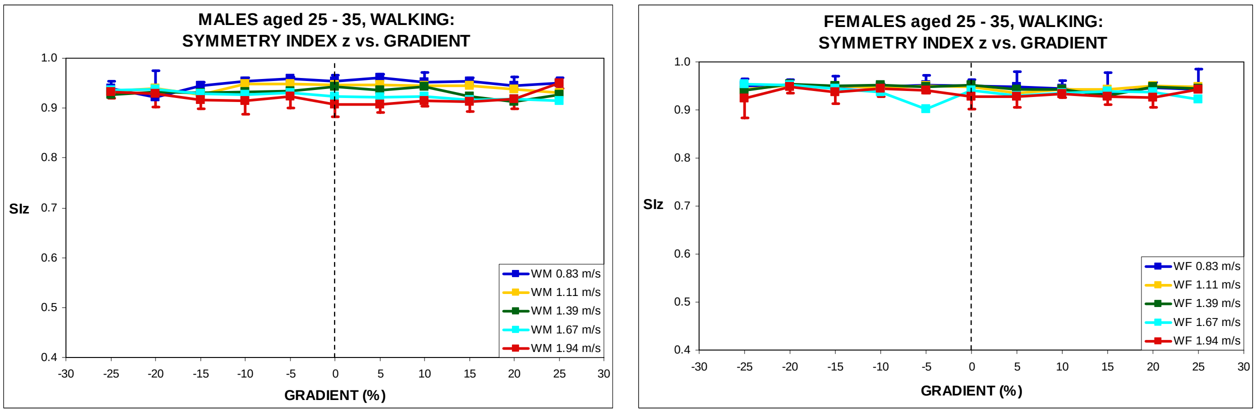

Figure 8.15. Symmetry Index in the medial/lateral direction as a function of gradient in walking, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index z is only little dependent on gradient;
- its patterns are quite similar at each speed;
- on average, females (0.942 ± 0.019, independently of speed and gradient) are slightly more symmetrical than males (0.933 ± 0.016);
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index.

In conclusion, we could state that, both in males and females, this direction is the most symmetrical one.

Specific results of the statistical analysis (with relevance) are shown in par. 6.3, Table 8.6c (see Appendix 8.1).

4.7. Symmetry Index as a function of gradient in running

A. In the **anterior/posterior** direction (Figure 8.16), our results show that:

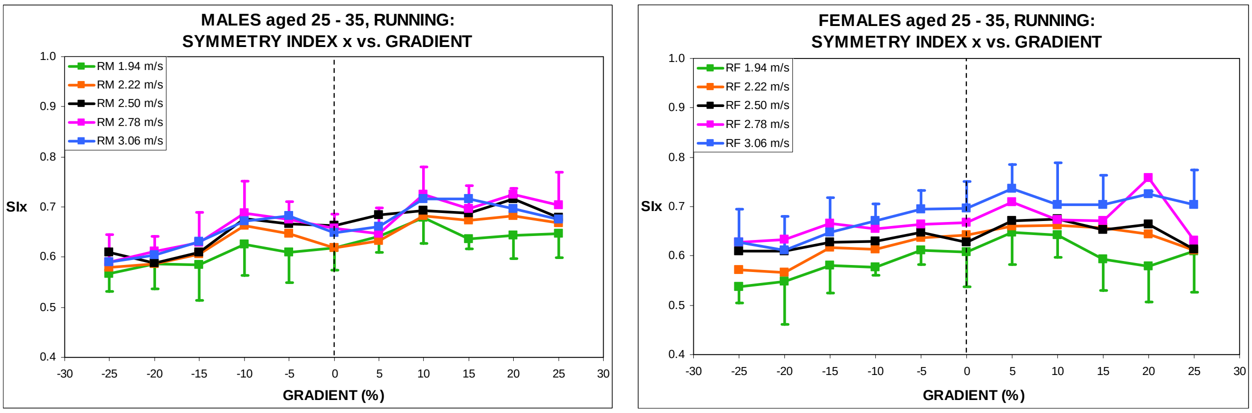

Figure 8.16. Symmetry Index in the anterior/posterior direction as a function of gradient in running, males (on the left) and females (on the right).

- both in males and females, Symmetry Index x is dependent on gradient;
- particularly, in males (left graph), it slightly increases with gradient up to -10% (from 0.587 ± 0.040 to 0.665 ± 0.065, at all the investigated speeds); from -10% to 0% it little decreases (from 0.665 ± 0.065 to 0.641 ± 0.036); from 0% to 10% it little increases (from 0.641 ± 0.036 to 0.699 ± 0.054); above this slope, it does not significantly change;
- however, in females (right graph), it slightly increases with gradient up to 5% (from 0.594 ± 0.052 to 0.685 ± 0.054); above this slope, it does not significantly change;
- its patterns are quite similar at each speed;
- on average, in the downhill gait, in males (0.623 ± 0.052, at all the investigated speeds) it is higher than the corresponding female values (from 0.619 ± 0.048);
- on average, in both the level and the uphill gaits, its values in males (0.641 ± 0.036 and 0.680 ± 0.048, respectively) are similar to female values (0.648 ± 0.063 and 0.664 ± 0.064);
- moreover, it is important to underline that the lower the speed, the lower the corresponding Symmetry Index.

In conclusion, we could state that, both in males and females, this represents the most asymmetrical movement direction.

Specific results of the statistical analysis (with relevance) are shown in par. 7.1, Table 8.7a (see Appendix 8.1).

B. In the **vertical** direction (Figure 8.17), our results show that:

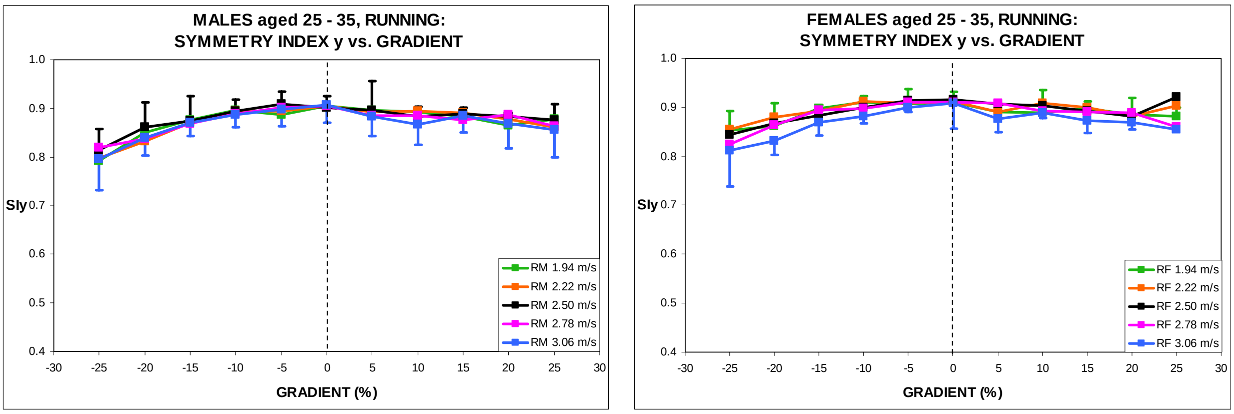

Figure 8.17. Symmetry Index in the vertical direction as a function of gradient in running, males (on the left) and females (on the right).

- both in males and females, Symmetry Index y is only slightly dependent on gradient;
- particularly, the qualitative analysis and the statistical one have shown that it increases up to the level condition (from 0.804 ± 0.067 to 0.905 ± 0.028, at all the investigated speeds, in

males; and from 0.838 ± 0.040 to 0.912 ± 0.010, in females); above this gradient, it does not significantly change;

- particularly, in males (left graph), the level gradient constitutes the most symmetrical condition (0.905 ± 0.028, at all the investigated speeds);
- in females (right graph), the level and the maximum up grades constitute the most symmetrical conditions (0.912 ± 0.010 and 0.885 ± 0.023, respectively);
- on average, its values in males (0.857 ± 0.042, at all the investigated speeds) are similar to female ones (0.864 ± 0.026);
- its patterns are similar at each speed;
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index.

In conclusion, we could state that, along this direction, there is a much more symmetry.

Specific results of the statistical analysis (with relevance) are shown in par. 7.2, Table 8.7b (see Appendix 8.1).

C. In the **medial/lateral** direction (Figure 8.18), our results show that:

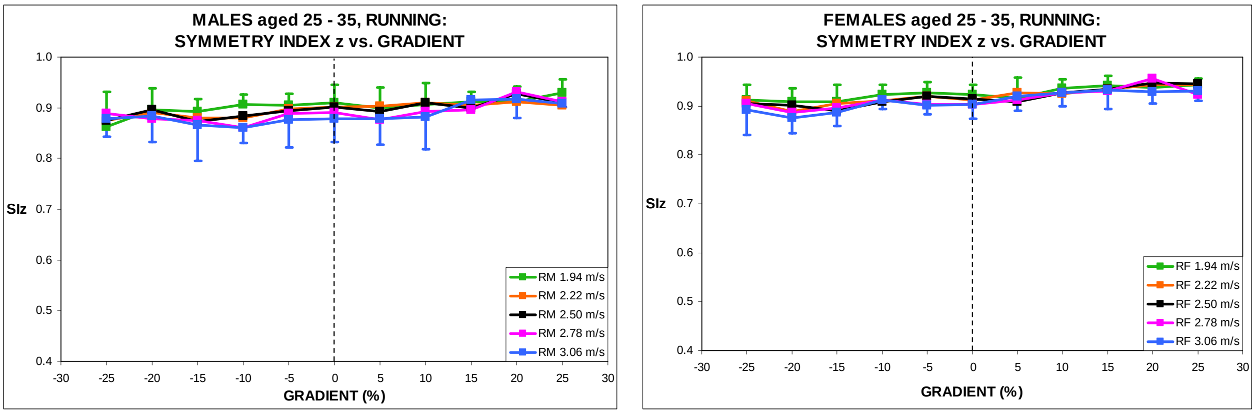

Figure 8.18. Symmetry Index in the medial/lateral direction as a function of gradient in running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), Symmetry Index z is only slightly dependent on gradient;
- particularly, the qualitative analysis and the statistical one have shown that Symmetry Index little increase with gradient (from 0.876 ± 0.048 to 0.913 ± 0.028, at all the investigated speeds, in males; and from 0.905 ± 0.035 to 0.937 ± 0.014, in females);
- both in males and females, the maximum up grade constitutes the most symmetrical condition (0.913 ± 0.028 and 0.937 ± 0.014, respectively);
- its patterns are similar at each speed;

- on average, its values in males (0.890 ± 0.039, at all the investigated speeds) are slightly lower than corresponding female values (0.913 ± 0.026). Furthermore, as already described, females are more symmetrical than males at the maximum up grades;
- moreover, it is important to underline that the higher the speed, the lower the corresponding Symmetry Index.

In conclusion, we could state that, both in males and females, this direction is the most symmetrical one.

Specific results of the statistical analysis (with relevance) are shown in par. 7.3, Table 8.7c (see Appendix 8.1).

5. CONCLUSION

The graphical (and qualitative) and the statistical analysis of Symmetry Index in each movement direction allow us to conclude that:

a) in each movement direction, Symmetry Index is slightly lower in young children (aged 6 to 13) and elderly adults (aged 56 to 65). This pattern is quite similar both in males and females, at each speed. As expected, this result clearly implies that young children and elderly adults are the most asymmetrical subjects: while during the early stages of the life this result could be ascribed to the process of gait maturation, asymmetries in old age are probably due to structural wearing down of the musculoskeletal system (Gabell et al., 1984; Minetti, 2006; Kang et al., 2008);

b) quite differently to what discussed in chapter 7 (par. 4), along the medial/lateral direction, there is the highest symmetry. This pattern occurs in each testing condition;

c) in level and gradient walking, along the anterior/posterior and vertical directions, the lowest is the speed, the lowest is the corresponding Symmetry Index. This pattern occurs in each testing condition. Knowing of this result seems to be important in partially questioning the problem related to the metabolic cost of walking in constant conditions (i.e. movement direction and speed; Minetti et al., 1993; 1997);

d) however, along the medial/lateral direction, the highest is the speed, the lowest is the corresponding Symmetry Index. This pattern occurs in each testing condition. It seems that such a result could be related to the positive ballistic effect;

e) moreover, in level and gradient running, along the forward direction, the lowest is the speed, the lowest is the corresponding Symmetry Index. This pattern occurs in each testing condition. As shown in walking, this result seems to be important in partially questioning the problem related to the metabolic cost of running in constant conditions (i.e. movement direction and speed; Minetti et al., 1993; 1994; 1997);

f) however, along the vertical and medial/lateral directions, the highest is the speed, the lowest is the corresponding Symmetry Index. This pattern occurs in each testing condition.

6. MEAN OVERALL SYMMETRY INDEX

6.1. Introduction

We have also calculated the mean overall Symmetry Index in order to complete the analysis of symmetry/asymmetry. Mean overall Symmetry Index represents a synthetic index that summarizes the results obtained separately in each movement direction. Particularly, it has been calculated as:

$$\text{Mean overall Symmetry Index} = \text{Average}\ (\text{SIx};\text{SIy};\text{SIz}) \quad [\text{Eq. } 8.4]$$

where *SIx* represents the Symmetry Index along the anterior/posterior direction; *SIy* the Symmetry Index along the vertical direction; and *SIz* the Symmetry Index along the medial/lateral one.

Single values of the mean overall Symmetry Index (males and females in all age group) are contained in the enclosed CD (First Study, Chapter 8, Mean overall Symmetry Index in level and gradient gaits).

6.2. Statistical analysis

Results will be presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The same statistical analyses described in par. 2 above have been carried out. Specifically, the statistical analysis as a function of gradient has been applied only for walking condition. Indeed, in running, gradients from -25% to 15% were considered for males, and from -25% to 0% for females.

Specifically, single values of the mean overall Symmetry Index are contained in the enclosed CD (First Study, Chapter 8, Statistical analysis).

Specific results of this statistical analysis are presented and described in the Appendix 8.1.

The graph legend is the same described in par. 3 above.

6.3. Mean overall Symmetry Index as a function of age

A. As a function of age, in males, our results show that:

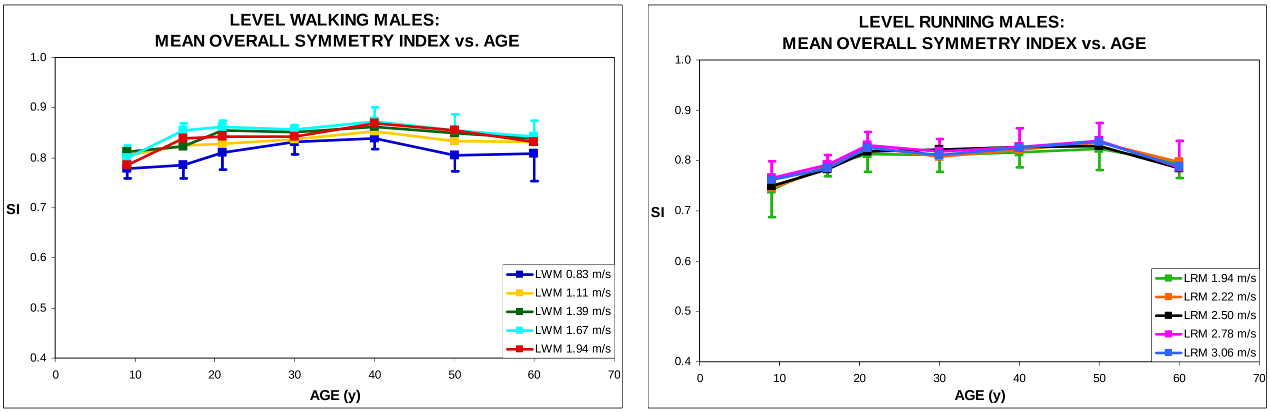

Figure 8.19. Mean overall Symmetry Index as a function of age in walking (on the left) and running (on the right), males.

- in level walking (left graph), mean overall Symmetry Index is lower in young children aged 6 to 13 (0.796 ± 0.035, at all the investigated speeds, $p<0.001$);
- in each age group, symmetry slightly increases with speed;
- in level running (right graph), mean overall Symmetry Index is lower in young children aged 6 to 17 (0.770 ± 0.036, at all the investigated speeds) and in elderly adults aged 56 to 65 (0.791 ± 0.042, at all the investigated speeds, $p<0.001$);
- in each age group, there are no significant differences among speeds.

B. Furthermore, in females:

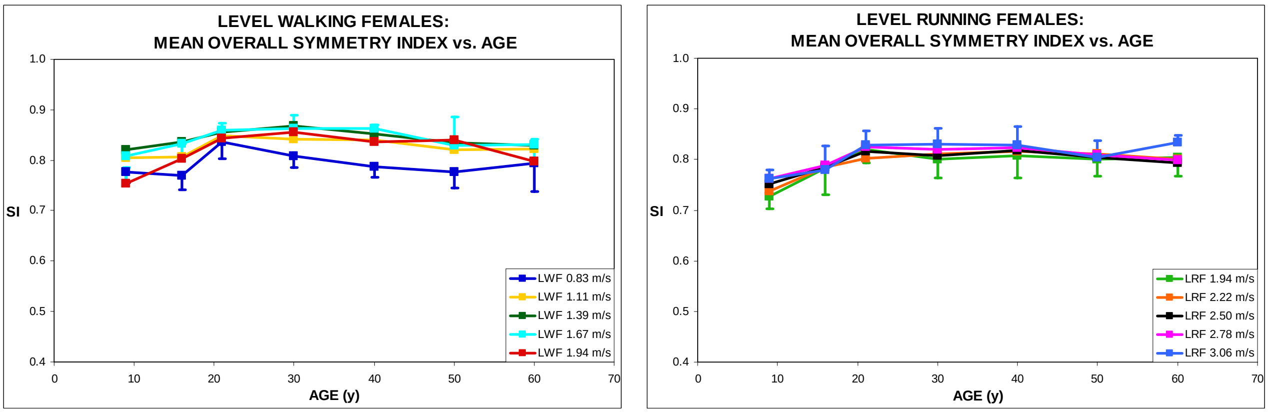

Figure 8.20. Mean overall Symmetry Index as a function of age in walking (on the left) and running (on the right), females.

- as shown in males, in level walking (left graph), mean overall Symmetry Index is lower in young children aged 6 to 13 (0.792 ± 0.031, at all the investigated speeds, $p<0.05$);
- in each age group, symmetry slightly increases with speed up to 1.39 m/s;
- in running (right graph), mean overall Symmetry Index is lower in young children aged 6 to 17 (0.775 ± 0.033, at all the investigated speeds, $p<0.001$);

- in each age group, there are no significant differences between each speed.

Specific results of the statistical analysis (with relevance) are shown in par. 7.3, Table 8.8a and 8.8c (see Appendix 8.1).

6.4. Mean overall Symmetry Index as a function of speed

A. As a function of speed, in males, our results show that:

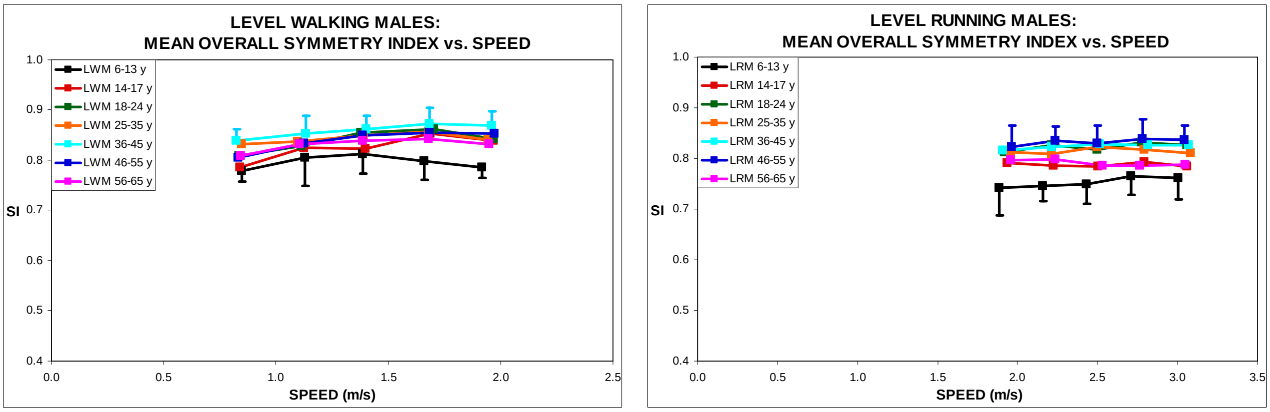

Figure 8.21. Mean overall Symmetry Index as a function of speed in walking (on the left) and running (on the right), males.

- on average, level walking (left graph: 0.833 ± 0.036, independently of age and speed) is more symmetrical than level running (right graph: 0.803 ± 0.036);
- in walking, mean overall Symmetry Index slightly increases with speeds in the 0.83 to 1.67 m/s range (from 0.808 ± 0.032 to 0.841 ± 0.039, independently of age, $p<0.05$); above this speed, it does not significantly change;
- as expected, children aged 6 to 13 are the less symmetrical subjects; otherwise, adults aged 36 to 45 are the most symmetrical subjects;
- in running, mean overall Symmetry Index does not change with speed;
- as expected, children aged 6 to 13 are the less symmetrical subjects; otherwise, adults aged 46 to 55 are the most symmetrical subjects.

B. Furthermore, in females:

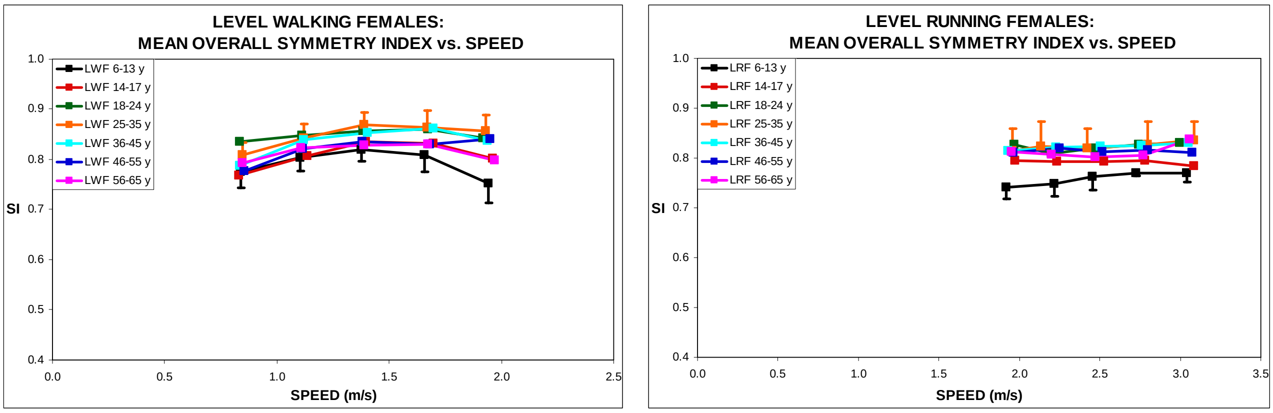

Figure 8.22. Mean overall Symmetry Index as a function of speed
in walking (on the left) and running (on the right), females.

- on average, level walking (left graph: 0.824 ± 0.037, independently of age and speed) is more symmetrical than level running (right graph: 0.807 ± 0.035);
- in walking, mean overall Symmetry Index slightly increases with speeds in the 0.83 to 1.67 m/s range (from 0.793 ± 0.039 to 0.841 ± 0.035, $p<0.01$); above this speed, it slightly decreases ($p<0.001$);
- as expected, children aged 6 to 13 and elderly adults aged 56 to 65 are the less symmetrical subjects; otherwise, young adults aged 25 to 35 are the most symmetrical subjects;
- in running, mean overall Symmetry Index does not change with speed.

Specific results of the statistical analysis (with relevance) are shown in par. 7.3, Table 8.8b and 8.8d (see Appendix 8.1).

6.5. Mean overall Symmetry Index as a function of gradient

A. As a function of gradient, in males aged 25 to 35, our results show that:

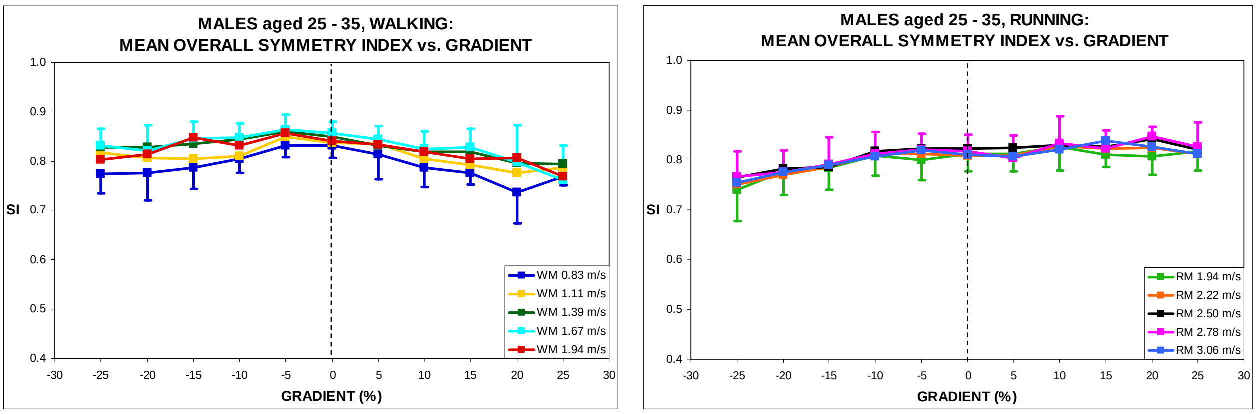

Figure 8.23. Mean overall Symmetry Index as a function of gradient
in walking (on the left) and running (on the right), males.

- both in walking (left graph) and running (right graph), the lower the speed, the lower the mean overall Symmetry Index;
- on average, gradient walking (0.816 ± 0.038, independently of speed and gradient) is more symmetrical than gradient running (0.806 ± 0.042);
- at all the investigated speeds, in gradient walking, mean overall Symmetry Index does not change up to 0%; above this slope, it slightly decreases (from 0.843 ± 0.080 to 0.776 ± 0.041, $p<0.05$);
- however, at all the investigated speeds, in gradient running, mean overall Symmetry Index slightly increases up to -10% (from 0.811 ± 0.034 to 0.828 ± 0.038, $p<0.05$); it then does not significantly change.

B. Furthermore, in females aged 25 to 35:

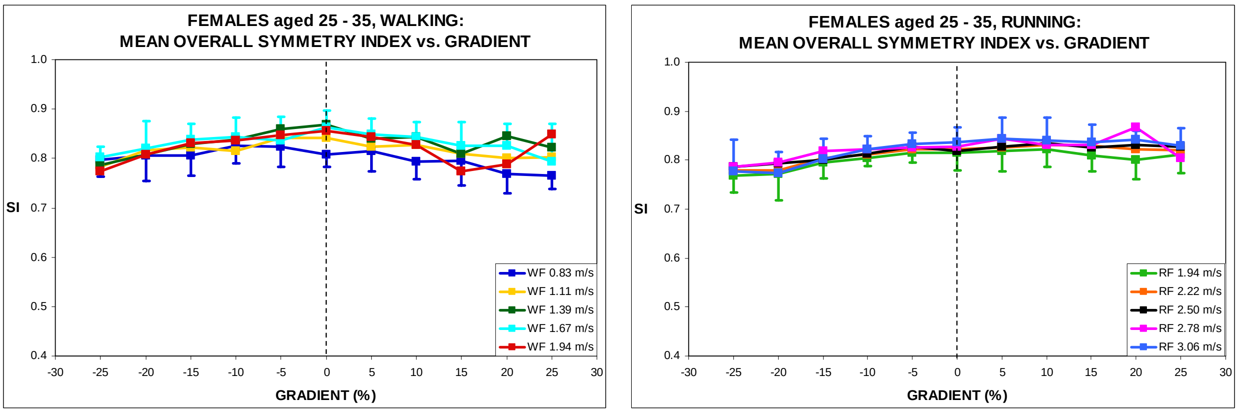

Figure 8.24. Mean overall Symmetry Index as a function of gradient in walking (on the left) and running (on the right), females.

- both in walking (left graph) and running (right graph), the lower the speed, the lower the mean overall Symmetry Index;
- on average, gradient walking (0.820 ± 0.041, independently of speed and gradient) is slightly more symmetrical than gradient running (0.815 ± 0.034);
- at all the investigated speeds, in gradient walking, mean overall Symmetry Index slightly increases up to 0% (from 0.789 ± 0.042 to 0.848 ± 0.029, $p<0.05$); above this slope, it slightly decreases;
- however, in gradient running, mean overall Symmetry Index slightly increases up to -10% (from 0.779 ± 0.042 to 0.814 ± 0.022, $p<0.05$); it then does not significantly change.

Specific results of the statistical analysis (with relevance) are shown in par. 7.3, Table 8.8e and 8.8f (see Appendix 8.1).

6.6. Conclusion

To sum up, our results show that, independently of movement direction:

a) young male and female children (aged 6 to 13) and elderly female adults (aged 56 to 65) seem to be the most asymmetrical age subject groups. Other than that, subjects coming from the other age groups (aged 14 to 55) seem to be more symmetrical;

b) in level walking, symmetry slightly increases with speed. This pattern could be related to the positive ballistic effect. However, this pattern does not occur in level running. Probably, this could be due to the highest metabolic cost observed in constant conditions of running;

c) in gradient walking, males' symmetry increases with slope. This highlights the fact that there could be a major motor control necessary to maintain a higher moving force (Jordan et al., 2008). However, this pattern appears to be inverted in females in whom a lower force is related to a minor motor control;

d) moreover, in gradient running, males' and females' symmetry slightly increases with slope. As expected, this result confirms that running is more dependent on both agility and force in comparison to walking (Minetti et al., 1994).

Areas for further research include similar analysis for other age groups, like younger subjects (< 6 years) for their different locomotion techniques and lower limb length and older ones (> 65 years) for their articular constraints.

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Appendix 8.1

SYMMETRY INDEX: STATISTICAL ANALYSIS

1. INTRODUCTION

In this appendix, there are cross-section tables showing statistical results (with relevances) in Symmetry Index (SI) in each testing condition. They are arranged in the same order as already presented in chapter 8.

As previously described, the alpha test level set for statistical significance was 0.05. If there is no significance (p=NS), the corresponding space in empty. In all the other cases, the significance is specified: in blue for males and in pink for females.

2. LEVEL WALKING: SYMMETRY INDEX AS A FUNCTION OF AGE

2.1. Anterior/posterior direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a low significance in Symmetry Index x as a function of age, in relation to each speed.

LEVEL WALKING	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**		$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$	$p<0.01$		
14 - 17 y		**Males** **Females**					
18 - 24 y	$p<0.05$ $p<0.05$		**Males** **Females**				$p<0.05$
25 - 35 y	$p<0.05$ $p<0.05$			**Males** **Females**			
36 - 45 y	$p<0.01$				**Males** **Females**		
46 - 55 y						**Males** **Females**	
56 - 65 y			$p<0.05$				**Males** **Females**

Table 8.2a. SI x as a function of age in level walking (males and females).

2.2. Vertical direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a significance in Symmetry Index y as a function of age, in relation to each speed.

LEVEL WALKING	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**	$p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.05$	$p<0.001$	$p<0.001$	$p<0.001$
14 - 17 y	$p<0.001$	**Males** **Females**	$p<0.01$	$p<0.01$			
18 - 24 y	$p<0.001$ $p<0.01$	$p<0.01$	**Males** **Females**			$p<0.05$	$p<0.05$
25 - 35 y	$p<0.001$ $p<0.05$	$p<0.01$		**Males** **Females**		$p<0.05$	$p<0.05$
36 - 45 y	$p<0.001$				**Males** **Females**		$p<0.05$
46 - 55 y	$p<0.001$		$p<0.05$	$p<0.05$		**Males** **Females**	
56 - 65 y	$p<0.001$		$p<0.05$	$p<0.05$	$p<0.05$		**Males** **Females**

Table 8.2b. SI y as a function of age in level walking (males and females).

2.3. Medial/lateral direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a low significance in Symmetry Index z as a function of age, in relation to each speed.

LEVEL WALKING	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**		$p<0.05$	$p<0.05$	$p<0.01$	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.01$
14 - 17 y		**Males** **Females**				$p<0.05$ $p<0.05$	
18 - 24 y	$p<0.05$		**Males** **Females**				
25 - 35 y	$p<0.05$			**Males** **Females**			
36 - 45 y	$p<0.01$				**Males** **Females**		
46 -55 y	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.05$				**Males** **Females**	
56 - 65 y	$p<0.05$ $p<0.01$						**Males** **Females**

Table 8.2c. SI z as a function of age in level walking (males and females).

3. LEVEL RUNNING: SYMMETRY INDEX AS A FUNCTION OF AGE

3.1. Anterior/posterior direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a high significance in Symmetry Index x as a function of age, in relation to each speed.

LEVEL RUNNING	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**	$p<0.05$ $p<0.05$	$p<0.001$ $p<0.001$	$p<0.01$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.01$	$p<0.05$ $p<0.05$
14 - 17 y	$p<0.05$ $p<0.05$	**Males** **Females**	$p<0.05$ $p<0.01$		$p<0.01$ $p<0.001$	$p<0.01$	
18 -24 y	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.01$	**Males** **Females**	$p<0.05$ $p<0.05$		 $p<0.05$	$p<0.01$
25 - 35 y	$p<0.01$		$p<0.05$ $p<0.05$	**Males** **Females**	$p<0.05$ $p<0.05$		
36 - 45 y	$p<0.001$ $p<0.001$	$p<0.01$ $p<0.001$		$p<0.05$ $p<0.05$	**Males** **Females**	 $p<0.05$	$p<0.01$
46 - 55 y	$p<0.001$ $p<0.01$	$p<0.01$	 $p<0.05$		 $p<0.05$	**Males** **Females**	$p<0.01$
56 - 65 y	$p<0.05$ $p<0.05$		$p<0.01$		$p<0.01$	$p<0.01$	**Males** **Females**

Table 8.3a. SI x as a function of age in level running (males and females).

3.2. Vertical direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a high significance in Symmetry Index y as a function of age, in relation to each speed.

LEVEL RUNNING	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
14 - 17 y	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.05$ $p<0.01$	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$
18 - 24 y	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.01$	**Males** **Females**	 $p<0.001$	$p<0.05$ $p<0.01$	$p<0.01$ $p<0.001$	$p<0.001$ $p<0.001$
25 - 35 y	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	 $p<0.001$	**Males** **Females**	 $p<0.001$	$p<0.01$ $p<0.001$	$p<0.001$ $p<0.001$
36 - 45 y	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.05$ $p<0.01$	 $p<0.001$	**Males** **Females**	 $p<0.05$	$p<0.001$
46 - 55 y	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.01$ $p<0.001$	$p<0.01$ $p<0.001$	 $p<0.05$	**Males** **Females**	$p<0.001$ $p<0.05$
56 - 65 y	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.001$ $p<0.05$	**Males** **Females**

Table 8.3b. SI y as a function of age in level running (males and females).

3.3. Medial/lateral direction

Both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a high significance in Symmetry Index z as a function of age, in relation to each speed.

LEVEL RUNNING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males** **Females**		p<0.001	p<0.001 p<0.001	p<0.05 p<0.05	p<0.001 p<0.001	p<0.01
14 - 17 y		**Males** **Females**	p<0.001	p<0.001 p<0.001	p<0.001	p<0.001 p<0.001	p<0.05 p<0.05
18 - 24 y	p<0.001	p<0.001	**Males** **Females**	p<0.001	p<0.01	p<0.001	
25 - 35 y	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001	**Males** **Females**	p<0.05 p<0.001		p<0.01
36 - 45 y	p<0.05 p<0.05	p<0.001	p<0.01	p<0.05 p<0.001	**Males** **Females**	p<0.05 p<0.001	
46 - 55 y	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001		p<0.05 p<0.001	**Males** **Females**	p<0.01
56 - 65 y	p<0.01	p<0.05 p<0.05		p<0.01		p<0.01	**Males** **Females**

Table 8.3c. SI z as a function of age in level running (males and females).

4. LEVEL WALKING: SYMMETRY INDEX AS A FUNCTION OF SPEED

4.1. Anterior/posterior direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that walking is highly different to running (p<0.001).

Moreover, both in males and females, there is a high significance as a function of speed in each age group (p<0.001). This pattern occurs up to the speed of 1.39 m/s.

LEVEL WALKING in all males and in females aged 18 to 55	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Males** **Females**	p<0.01 p<0.001	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.001
1.11 m/s	p<0.01 p<0.001	**Males** **Females**	p<0.05 p<0.001	p<0.01 p<0.01	p<0.05 p<0.01
1.39 m/s	p<0.001 p<0.001	p<0.05 p<0.001	**Males** **Females**		
1.67 m/s	p<0.001 p<0.001	p<0.01 p<0.01		**Males** **Females**	
1.94 m/s	p<0.001 p<0.001	p<0.05 p<0.01			**Males** **Females**

Table 8.4a. SI x as a function of speed in level walking (males and females aged 18 to 55).

LEVEL WALKING in females aged 6 to 17 and 56 to 65	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Females**	p<0.001	p<0.001	p<0.001	p<0.001
1.11 m/s	p<0.001	**Females**	p<0.001	p<0.01	p<0.01
1.39 m/s	p<0.001	p<0.001	**Females**		
1.67 m/s	p<0.001	p<0.01		**Females**	p<0.01 and p<0.05
1.94 m/s	p<0.001	p<0.01		p<0.01 and p<0.05	**Females**

Table 8.4b. SI x as a function of speed in level walking (females aged 6 to 17 and 56 to 65).

4.2. Vertical direction

Both in males and females, there is a high significance as a function of speed in each age group (p<0.001). This pattern occurs up to the speed of 1.11 m/s.

LEVEL WALKING	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
1.11 m/s	$p<0.001$ $p<0.001$	**Males** **Females**			
1.39 m/s	$p<0.001$ $p<0.001$		**Males** **Females**		
1.67 m/s	$p<0.001$ $p<0.001$			**Males** **Females**	
1.94 m/s	$p<0.001$ $p<0.001$				**Males** **Females**

Table 8.4c. SI y as a function of speed in level walking (males and females).

LEVEL WALKING in subjects aged 6 to 17 and in females aged 36 to 45	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
1.11 m/s	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.05$ $p<0.05$		
1.39 m/s	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.05$	**Males** **Females**		
1.67 m/s	$p<0.001$ $p<0.001$			**Males** **Females**	
1.94 m/s	$p<0.001$ $p<0.001$				**Males** **Females**

Table 8.4d. SI y as a function of speed in level walking

(subjects aged 6 to 17 and females aged 36 to 45).

4.3. Medial/lateral direction

Both in males and females, there is a high significance as a function of speed in each age group ($p<0.001$).

LEVEL WALKING	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Males** **Females**	$p<0.01$	$p<0.01$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$
1.11 m/s	$p<0.01$	**Males** **Females**	$p<0.05$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$
1.39 m/s	$p<0.01$	$p<0.05$	**Males** **Females**	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$
1.67 m/s	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	**Males** **Females**	$p<0.001$ $p<0.01$
1.94 m/s	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	**Males** **Females**

Table 8.4e. SI z as a function of speed in level walking (males and females).

5. LEVEL RUNNING: SYMMETRY INDEX AS A FUNCTION OF SPEED

5.1. Anterior/posterior direction

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that:

- in male and female young subjects aged 6 to 13, a high significance in Symmetry Index x as a function of speed (p<0.001);

SUBJECTS	LEVEL RUNNING in subjects aged 6 to 13
Males	p<0.001 comparing all speeds
Females	p<0.001 comparing all speeds

Table 8.5a. SI x as a function of speed in level running (subjects aged 6 to 13).

- in males and females aged 14 to 55, a high significance in Symmetry Index x up to 1.67 m/s (p<0.01);

SUBJECTS	LEVEL RUNNING in subjects aged 14 to 55
Males	p<0.01 from 0.83 to 1.67 m/s
Females	p<0.01 from 0.83 to 1.67 m/s

Table 8.5b. SI x as a function of speed in level running (subjects aged 14 to 55).

- in addition, in male and female elderly adults aged 56 to 65, Symmetry Index x does not significantly change with speed.

SUBJECTS	LEVEL RUNNING in subjects aged 56 to 65
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 8.5c. SI x as a function of speed in level running (subjects aged 56 to 65).

5.2. Vertical direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show no significance in Symmetry Index y as a function of speed.

The only exception is in females aged 46 to 55. In this case, Symmetry Index y slightly decreases with speed (p<0.01).

SUBJECTS	LEVEL RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 8.5d. SI y as a function of speed in level running (males and females).

5.3. Medial/lateral direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that Symmetry Index z slightly decreases with speed ($p<0.01$). This pattern is similar in all age groups.

SUBJECTS	**LEVEL RUNNING in each age group**
Males	$p<0.01$ comparing all speeds
Females	$p<0.01$ comparing all speeds

Table 8.5e. SI z as a function of speed in level running (males and females).

6. WALKING: SYMMETRY INDEX AS A FUNCTION OF GRADIENT

6.1. Anterior/posterior direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a high significance in Symmetry Index x as a function of gradient, at each speed.

GRADIENT WALKING	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.05$	$p<0.05$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.01$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.01$
-20%	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.05$ $p<0.05$	$p<0.01$ $p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.05$	$p<0.05$ $p<0.01$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$
-15%	$p<0.05$	$p<0.05$ $p<0.05$	**Males** **Females**		$p<0.05$ $p<0.01$	$p<0.05$ $p<0.05$	$p<0.01$	$p<0.05$	$p<0.001$	$p<0.01$	$p<0.05$
-10%	$p<0.05$ $p<0.001$	$p<0.01$ $p<0.001$		**Males** **Females**	$p<0.01$ $p<0.05$	$p<0.01$			$p<0.01$ $p<0.05$	$p<0.05$	$p<0.01$
-5%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.01$	$p<0.05$ $p<0.01$	$p<0.01$ $p<0.05$	**Males** **Females**		$p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	$p<0.01$ $p<0.05$	$p<0.001$
0%	$p<0.05$ $p<0.05$	$p<0.001$ $p<0.05$	$p<0.05$ $p<0.05$	$p<0.01$		**Males** **Females**		$p<0.01$ $p<0.05$	$p<0.001$ $p<0.05$	$p<0.01$ $p<0.01$	$p<0.01$
5%	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.01$	$p<0.01$		$p<0.001$		**Males** **Females**	$p<0.001$	$p<0.01$ $p<0.05$	$p<0.01$ $p<0.05$	$p<0.01$
10%	$p<0.05$	$p<0.05$	$p<0.05$		$p<0.001$ $p<0.01$	$p<0.01$ $p<0.05$	$p<0.001$	**Males** **Females**	$p<0.05$	$p<0.05$	$p<0.05$ $p<0.05$
15%	$p<0.05$	$p<0.05$	$p<0.001$	$p<0.01$ $p<0.05$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.05$	$p<0.01$ $p<0.05$	$p<0.05$	**Males** **Females**		
20%	$p<0.05$	$p<0.05$	$p<0.01$	$p<0.05$	$p<0.01$ $p<0.05$	$p<0.01$ $p<0.01$	$p<0.01$ $p<0.05$	$p<0.05$		**Males** **Females**	
25%	$p<0.01$	$p<0.05$	$p<0.05$	$p<0.01$	$p<0.001$	$p<0.01$	$p<0.01$	$p<0.05$ $p<0.05$			**Males** **Females**

Table 8.6a. SI x as a function of gradient in walking (males and females).

6.2. Vertical direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a slightly significance in Symmetry Index y as a function of gradient, at each speed.

GRADIENT WALKING	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**					$p<0.05$					
-20%		**Males** **Females**			$p<0.05$	$p<0.05$ $p<0.05$					
-15%			**Males** **Females**								$p<0.05$
-10%				**Males** **Females**							$p<0.05$
-5%		$p<0.05$			**Males** **Females**			$p<0.05$		$p<0.05$	$p<0.01$
0%	$p<0.05$	$p<0.05$ $p<0.05$				**Males** **Females**		$p<0.01$		$p<0.05$	$p<0.05$ $p<0.01$
5%							**Males** **Females**				$p<0.05$
10%					$p<0.05$	$p<0.01$		**Males** **Females**			$p<0.05$
15%									**Males** **Females**		
20%					$p<0.05$	$p<0.05$				**Males** **Females**	
25%			$p<0.05$	$p<0.05$	$p<0.01$	$p<0.05$ $p<0.01$	$p<0.05$	$p<0.05$			**Males** **Females**

Table 8.6b. SI y as a function of gradient in walking (males and females).

6.3. Medial/lateral direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a slightly significance in Symmetry Index z as a function of gradient, at each speed.

GRADIENT WALKING	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**										
-20%		**Males** **Females**					$p<0.05$	$p<0.01$	$p<0.01$		
-15%			**Males** **Females**		$p<0.05$		$p<0.05$	$p<0.001$	$p<0.05$		
-10%				**Males** **Females**	$p<0.05$		$p<0.05$	$p<0.05$			
-5%			$p<0.05$	$p<0.05$	**Males** **Females**						
0%						**Males** **Females**	$p<0.05$				
5%		$p<0.05$	$p<0.05$	$p<0.05$		$p<0.05$	**Males** **Females**				
10%		$p<0.01$	$p<0.001$	$p<0.05$				**Males** **Females**			
15%		$p<0.01$	$p<0.05$						**Males** **Females**		
20%										**Males** **Females**	
25%											**Males** **Females**

Table 8.6c. SI z as a function of gradient in walking (males and females).

7. RUNNING: SYMMETRY INDEX AS A FUNCTION OF GRADIENT

7.1. Anterior/posterior direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a high significance in Symmetry Index x as a function of gradient, at each speed.

GRADIENT RUNNING	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	**Males** **Females**		 p<0.01	p<0.001 p<0.001	p<0.01 p<0.01	p<0.001 p<0.01	p<0.001 p<0.001		p<0.001 p<0.01	p<0.001 p<0.05	p<0.001
-20%		**Males** **Females**	 p<0.01	p<0.001 p<0.001	p<0.01 p<0.01	p<0.01 p<0.01	p<0.01 p<0.001	p<0.001 p<0.01	p<0.001 p<0.05	p<0.001 p<0.05	p<0.001
-15%	 p<0.01	 p<0.01	**Males** **Females**	p<0.001	p<0.001 p<0.05	p<0.05	p<0.05 p<0.01	p<0.001 p<0.05	p<0.001 p<0.05	p<0.001	p<0.001
-10%	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001	**Males** **Females**	 p<0.05	p<0.05 p<0.05	 p<0.001	p<0.05 p<0.05	 p<0.05	p<0.01	
-5%	p<0.01 p<0.01	p<0.01 p<0.01	p<0.001 p<0.05	 p<0.05	**Males** **Females**		 p<0.01	p<0.01 p<0.05	p<0.001	p<0.01	
0%	p<0.001 p<0.01	p<0.01 p<0.01	p<0.05	p<0.05 p<0.05		**Males** **Females**	 p<0.05	p<0.001	p<0.05 p<0.05	p<0.01	p<0.05 p<0.01
5%	p<0.001 p<0.001	p<0.01 p<0.001	p<0.05 p<0.01	 p<0.001	 p<0.01	 p<0.05	**Males** **Females**	p<0.05		p<0.05	 p<0.05
10%		p<0.001 p<0.01	p<0.001 p<0.05	p<0.05 p<0.05	p<0.01 p<0.05	p<0.001	p<0.05	**Males** **Females**			p<0.05
15%	p<0.001 p<0.01	p<0.001 p<0.05	p<0.001 p<0.05	 p<0.05	p<0.001	p<0.05 p<0.05			**Males** **Females**		
20%	p<0.001 p<0.05	p<0.001 p<0.05	p<0.001	p<0.01	p<0.01	p<0.01	p<0.05			**Males** **Females**	
25%	p<0.001	p<0.001	p<0.001			p<0.05 p<0.01	 p<0.05	p<0.05			**Males** **Females**

Table 8.7a. SI x as a function of gradient in running (males and females).

7.2. Vertical direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a high significance in Symmetry Index y as a function of gradient, at each speed.

GRADIENT RUNNING	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	**Males** **Females**	p<0.01 p<0.05	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.01	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.01	p<0.001 p<0.05
-20%	p<0.01 p<0.05	**Males** **Females**	p<0.001 p<0.05	p<0.001 p<0.001	p<0.001 p<0.01	p<0.001 p<0.001	p<0.001 p<0.05	p<0.01 p<0.01	p<0.01 p<0.001	p<0.05 p<0.05	p<0.01
-15%	p<0.001 p<0.001	p<0.001 p<0.05	**Males** **Females**	p<0.001 p<0.05	p<0.01 p<0.05	p<0.001 p<0.001	p<0.001	p<0.05	p<0.05	 p<0.05	
-10%	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.05	**Males** **Females**		p<0.01		 p<0.05	p<0.05 p<0.05	p<0.01 p<0.05	p<0.01
-5%	p<0.001 p<0.001	p<0.001 p<0.01	p<0.01 p<0.05		**Males** **Females**			 p<0.05	p<0.05 p<0.05	p<0.001 p<0.001	p<0.01
0%	p<0.001 p<0.001	p<0.001 p<0.001	p<0.001 p<0.001	p<0.01		**Males** **Females**	p<0.05 p<0.05	p<0.05 p<0.01	p<0.01 p<0.05	p<0.01 p<0.001	p<0.001
5%	p<0.001 p<0.01	p<0.001 p<0.05	p<0.001			p<0.05 p<0.05	**Males** **Females**			p<0.05 p<0.05	p<0.001
10%	p<0.001 p<0.001	p<0.01 p<0.01	p<0.05	 p<0.05	 p<0.05	p<0.05 p<0.01		**Males** **Females**		 p<0.05	p<0.05
15%	p<0.001 p<0.001	p<0.01 p<0.001	p<0.05	p<0.05 p<0.05	p<0.05 p<0.05	p<0.01 p<0.05			**Males** **Females**	 p<0.05	p<0.05
20%	p<0.001 p<0.01	p<0.05 p<0.05	 p<0.05	p<0.01 p<0.05	p<0.001 p<0.001	p<0.01 p<0.001	p<0.05 p<0.05	 p<0.05	 p<0.05	**Males** **Females**	
25%	p<0.001 p<0.05	p<0.01		p<0.01	p<0.01	p<0.001	p<0.001	p<0.05	p<0.05		**Males** **Females**

Table 8.7b. SI y as a function of gradient in running (males and females).

7.3. Medial/lateral direction

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a significance in Symmetry Index z as a function of gradient, at each speed.

GRADIENT RUNNING	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	**Males** **Females**	$p<0.05$			$p<0.05$				$p<0.05$	$p<0.001$	$p<0.05$
-20%	$p<0.05$	**Males** **Females**					$p<0.05$	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.001$	$p<0.01$
-15%			**Males** **Females**		$p<0.05$	$p<0.01$	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.001$	$p<0.05$ $p<0.05$	$p<0.01$ $p<0.001$	$p<0.001$
-10%				**Males** **Females**		$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.01$
-5%	$p<0.05$		$p<0.05$		**Males** **Females**	$p<0.05$				$p<0.05$	$p<0.05$
0%			$p<0.01$	$p<0.05$	$p<0.05$	**Males** **Females**	$p<0.001$			$p<0.05$	$p<0.05$
5%		$p<0.05$	$p<0.05$ $p<0.05$	$p<0.05$		$p<0.001$	**Males** **Females**			$p<0.05$	$p<0.05$
10%		$p<0.05$ $p<0.01$	$p<0.05$ $p<0.001$	$p<0.05$				**Males** **Females**			
15%	$p<0.05$	$p<0.05$ $p<0.01$	$p<0.05$ $p<0.05$	$p<0.05$					**Males** **Females**		
20%	$p<0.001$	$p<0.05$ $p<0.001$	$p<0.01$ $p<0.001$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$			**Males** **Females**	
25%	$p<0.05$	$p<0.01$	$p<0.001$	$p<0.01$	$p<0.05$	$p<0.05$	$p<0.05$				**Males** **Females**

Table 8.7c. SI z as a function of gradient in running (males and females).

8. MEAN OVERALL SYMMETRY INDEX

8.1. Level walking

As a function of age, both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a significance in mean overall Symmetry Index at each speed.

LEVEL WALKING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males** **Females**	$p<0.001$	$p<0.001$ $p<0.05$	$p<0.001$ $p<0.05$	$p<0.001$	$p<0.001$	$p<0.01$
14 - 17 y	$p<0.001$	**Males** **Females**					
18 - 24 y	$p<0.001$ $p<0.05$		**Males** **Females**				
25 - 35 y	$p<0.001$ $p<0.05$			**Males** **Females**			
36 - 45 y	$p<0.001$				**Males** **Females**		
46 - 55 y	$p<0.001$					**Males** **Females**	
56 - 65 y	$p<0.01$						**Males** **Females**

Table 8.8a. Mean overall SI as a function of age in level walking (males and females).

As a function of speed, both in males and females, repeated measures with the post-*hoc* paired *t*-test (with Bonferroni correction) show that walking is highly different to running ($p<0.001$).

Moreover, both in males and females, there is a high significance in mean overall Symmetry Index in each age group ($p<0.001$).

SUBJECTS	LEVEL WALKING in each age group
Males	$p<0.05$ up to 1.67 m/s
Females	$p<0.01$ up to 1.67 m/s

Table 8.8b. Mean overall SI as a function of speed in level walking (males and females).

8.2. Level running

As a function of age, both in males and females, one-way ANOVA and the post-*hoc* Bonferroni test show a high significance in mean overall Symmetry Index at each speed.

LEVEL RUNNING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.01$ $p<0.001$
14 - 17 y	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$
18 - 24 y	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**			$p<0.05$	$p<0.001$
25 - 35 y	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$		**Males** **Females**	$p<0.05$	$p<0.001$ $p<0.05$	$p<0.001$
36 - 45 y	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$		$p<0.05$	**Males** **Females**	$p<0.05$ $p<0.05$	$p<0.001$
46 - 55 y	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$	$p<0.001$ $p<0.05$	$p<0.05$ $p<0.05$	**Males** **Females**	$p<0.001$
56 - 65 y	$p<0.01$ $p<0.001$	$p<0.05$	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	**Males** **Females**

Table 8.8c. Mean overall SI as a function of age in level running (males and females).

As a function of speed, both in males and females, there is no significance in mean overall Symmetry Index in each age group.

SUBJECTS	LEVEL RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 8.8d. Mean overall SI as a function of speed in level running (males and females).

8.3. Gradient walking

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a slightly significance in mean overall Symmetry Index as a function of gradient, independently of speed.

GRADIENT WALKING	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	Males Females	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$					
-20%	$p<0.05$	Males Females	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$					
-15%	$p<0.05$	$p<0.05$	Males Females	$p<0.05$	$p<0.05$	$p<0.05$					
-10%	$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$	$p<0.05$					
-5%	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$					
0%	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$
5%						$p<0.05$	Males Females		$p<0.05$	$p<0.05$	$p<0.05$
10%						$p<0.05$		Males Females	$p<0.05$	$p<0.05$	$p<0.05$
15%						$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$	$p<0.05$
20%						$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$
25%						$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females

Table 8.8e. Mean overall SI as a function of gradient in walking (males and females).

8.4. Gradient running

Both in males and females, one-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show a slightly significance in mean overall Symmetry Index as a function of gradient, independently of speed.

GRADIENT RUNNING	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	Males Females	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$							
-20%	$p<0.05$ $p<0.05$	Males Females	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$							
-15%	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$	Males Females	$p<0.05$ $p<0.05$							
-10%	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$	$p<0.05$ $p<0.05$	Males Females							
-5%					Males Females						
0%						Males Females	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$
5%						$p<0.05$	Males Females		$p<0.05$	$p<0.05$	$p<0.05$
10%						$p<0.05$		Males Females	$p<0.05$	$p<0.05$	$p<0.05$
15%						$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$	$p<0.05$
20%						$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females	$p<0.05$
25%						$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	Males Females

Table 8.8f. Mean overall SI as a function of gradient in running (males and females).

Chapter 9

POLAR LOGARITHM GRAPHS

1. INTRODUCTION

A. In a first step, Fourier analysis allowed us to mathematically describe and graphically illustrate the three-dimensional BCOM pattern by obtaining both amplitude (A) and phase (φ) coefficients from the first to the sixth Fourier Series (see chapters 6 and 7). These results have been achieved in each movement direction and testing condition.

B. In a second step, it becomes important to graphically represent how these coefficients simultaneously change as both the walking/running speed and the slope change, in males and females of different age groups. As a consequence, a polar logarithm graph could help us in reaching this important aim.

In this chapter, we will therefore focus on the main characteristics of a polar logarithm graph in order to better understand and represent our mathematical results.

2. POLAR LOGARITHM GRAPHS

2.1. Introduction: a brief history

The concepts of *angle* and *radius* were already used by ancient peoples of the 1st millennium BCE (i.e. by the astronomer Hipparchus, the mathematician Archimedes). In the 9th century CE, a Persian mathematician employed the use of spherical trigonometry and map projection methods in order to convert polar coordinates to a different coordinate system centered on a specific point on the sphere (Polar logarithm graph in Mathematics - Wikipedia, the free encyclopedia, 2009).

There are various accounts of the introduction of polar coordinates as part of a formal coordinate system. For instance: a) the book *Origin of Polar Coordinates* by the professor J.L. Coolidge's in Harvard University; b) the books of G. de Saint-Vincent and B. Cavalieri; c) the application of polar coordinates to calculate the length of parabolic arcs by B. Pascal; d) the book *Method of Fluxions* by I. Newton, in which he examined the transformations between polar coordinates and nine other coordinate systems; and e) the journal *Acta Eruditorum* by J. Bernoulli, in which he used a system with a point on a line, called the *pole* and *polar axis* respectively.

2.2. Polar coordinate system

Because of the circular nature of the polar coordinate system, many curves (i.e. circle: $r(\varphi) = 1$; polar rose: $r(\varphi) = 2\sin 4\varphi$; Archimedean spiral: $r(\varphi) = \varphi$ with $0 < \varphi < 6\pi$; Figure 9.1: the left, the middle and the right graph, respectively) could be described by a rather simple polar equation, whereas their Cartesian form is much more intricate (Polar coordinate system in Mathematics - Wikipedia, the free encyclopedia, 2009).

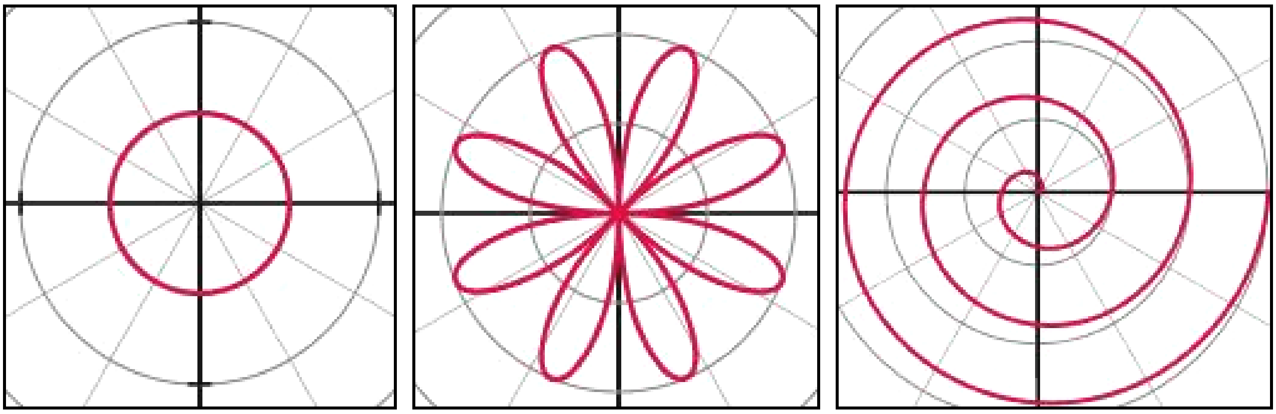


Figure 9.1. Some curves in a polar coordinate system.

The polar coordinate system could be extended into three dimensions with two different coordinate systems: 1) the cylindrical coordinate systems (left figure), and 2) the spherical coordinate systems (right figure), both of which include two-dimensional or planar polar coordinates as a subset. In essence, the cylindrical coordinate system extends polar coordinates by adding an additional distance coordinate, while the spherical system instead adds an additional angular coordinate (Figure 9.2).

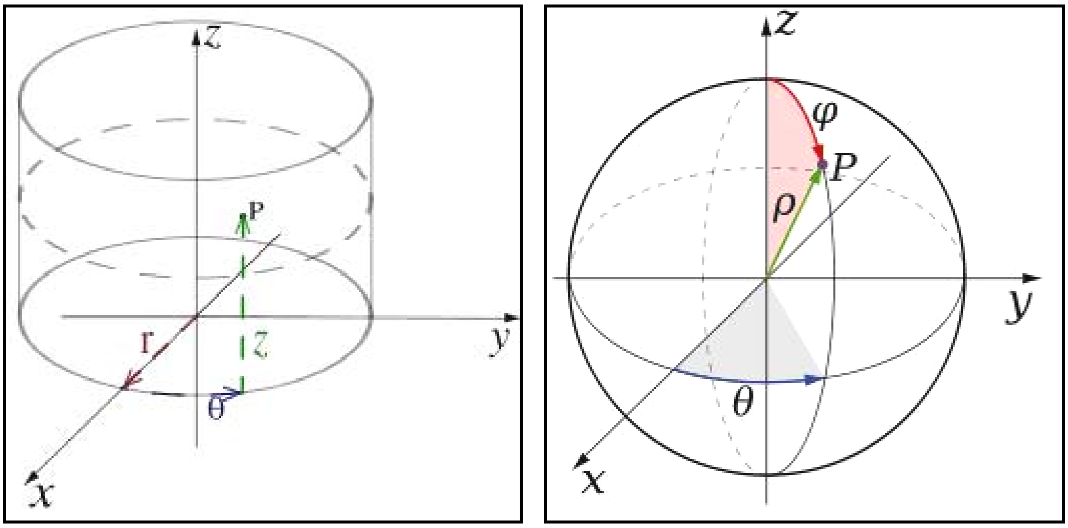


Figure 9.2. Cylindrical (on the left) and spherical (on the right) coordinate system.

The **polar coordinate system** is therefore a two-dimensional coordinate system in which each point on a plane is determined by a distance and an angle (Batschelet, 1975; 1981). This system is especially useful in situations where the relationship between two points is most easily expressed in terms of distances and angles; in the more familiar Cartesian or rectangular coordinate system, such a relationship can only be found through trigonometric formulation (see par. 2.3 below).

The equation defining an algebraic curve expressed in polar coordinates is known as a *polar equation*. In many cases, such an equation can simply be specified by defining A (distance or amplitude) as a function of φ (angle or phase). The resulting curve then consists of points of the form $(A(\varphi); \varphi)$ and can be regarded as the polar function A.

Different forms of symmetry could be deduced from the equation of a polar function A. For instance: a) if $A(-\varphi) = A(\varphi)$, the curve will be symmetrical about the horizontal (0°/180°) ray; b) if $A(\pi - \varphi) = A(\varphi)$, it will be symmetrical about the vertical (90°/270°) ray; and c) if $A(\varphi - \alpha°) = A(\varphi)$, it will be rotationally symmetrical $\alpha°$ counterclockwise about the pole.

2.3. Polar coordinates

Although Cartesian coordinates (bi-dimensional and three-dimensional) are used most often in biomechanics research, for some applications it is more convenient to use polar coordinates. Indeed, in Mathematics, 'the use of polar coordinates constitutes an alternative way to Cartesian coordinates' (Robertson et al., 2004).

The actual term *polar coordinates* has been attributed to G. Fontana and was used by 18th century. Particularly, in the polar system, as with the Cartesian system, two degrees of freedom describe the planar position of a point in a plane (Figure 9.3).

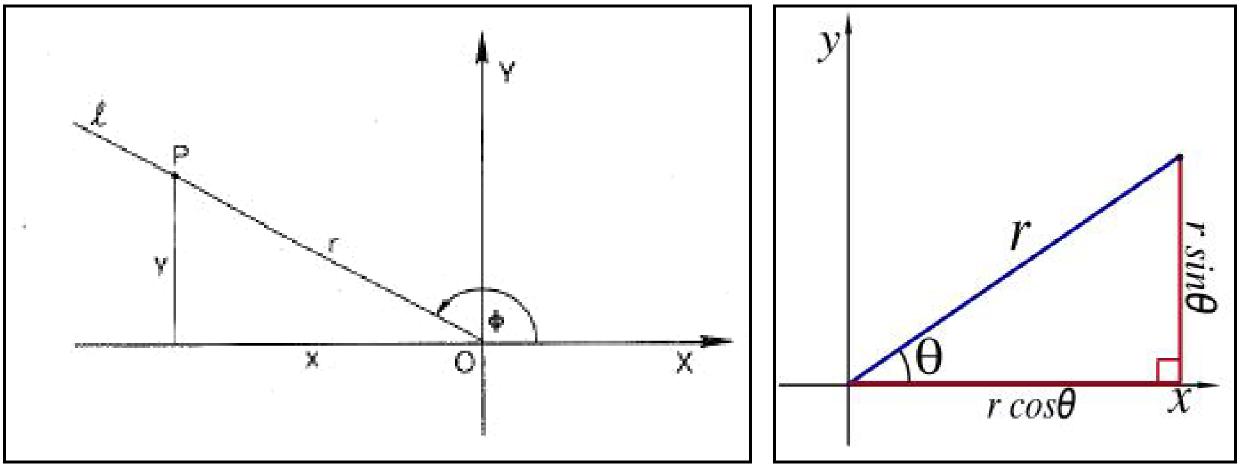


Figure 9.3. Rectangular (x and y, on the left) and polar (r or A, and φ, on the right) coordinates of a point P, in Batschelet (1981).

To be more specific:

- the **rectangular coordinates** of the point P are x and y, respectively (Batschelet, 1975; 1981; Robertson et al., 2004);

- the **polar coordinates** of the point P are A (or r, in the image above) and φ, respectively (Batschelet, 1975; 1981; Robertson et al., 2004). Indeed: **a)** the radial coordinate A denotes the point's distance from a central point known as the *pole* (equivalent to the *origin* in the Cartesian system); **b)** the angular coordinate (also known as the polar angle or the azimuth angle, and usually indicated by φ) denotes the positive or anticlockwise (counterclockwise) angle required to reach the point from the 0° ray or *polar axis* (which is equivalent to the *positive x-axis* in the Cartesian coordinate plane).

As shown in Figure 9.3 (above) and 9.4 (below), there is only one exceptional point, the origin O. It is given by A = 0 only, since φ is not defined (Batschelet, 1981).

Using simple trigonometry (Equation [9.1a] and [9.1b]), the general mathematical conversion from polar to Cartesian coordinates is accomplished as follows:

$$x = A \bullet \cos(\varphi) \quad \text{[Eq. 9.1a]}$$

$$y = A \bullet \sin(\varphi) \quad \text{[Eq. 9.1b]}$$

Furthermore, if the domain of φ consists of all real numbers, x and y are periodic functions of φ, since any new rotation around the unit circle (Batschelet, 1975) generates the same values of x and y (Batschelet, 1981; Equation [9.1c] and [9.1d]). The period is 360° or 2π:

$$x = \cos(\varphi) \quad \text{[Eq. 9.1c]}$$

$$y = \sin(\varphi) \quad \text{[Eq. 9.1d]}$$

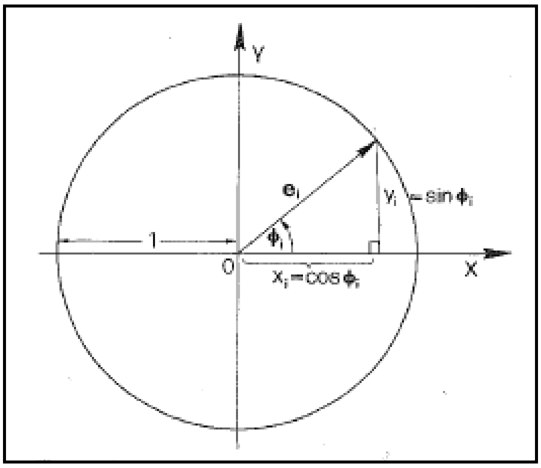


Figure 9.4. The rectangular coordinates of a unit vector, in Batschelet (1981).

In addition, it is also possible, given the Cartesian coordinates (x, y) of a point, to compute its polar coordinates (A, φ), using the following trigonometric relationships:

by a simple application of the Pythagorean theorem, $A = \sqrt{x^2 + y^2}$ [Eq. 9.1e]

where:

- *x* could be expressed as $\frac{1}{n} \bullet (\cos\varphi_1 + \cos\varphi_2 + ... + \cos\varphi_n)$;
- *y* could be expressed as $\frac{1}{n} \bullet (\sin\varphi_1 + \sin\varphi_2 + ... + \sin\varphi_n)$.

Consequently, the general Equation [9.1e] could be changed as:

$$A = \frac{1}{n} \bullet [(\Sigma \cos\varphi_1)^2 + (\Sigma \sin\varphi_1)^2] \text{ [Eq. 9.1f]}$$

To determine the angular coordinate φ, the following two ideas must be considered:

1. for A = 0, φ can be set to any real value;
2. for A ≠ 0, to get a unique representation for φ, it must be limited to an interval of size 2π. As a result, conventional choices for such an interval are [(0; 2π) and (-π; π)].

Therefore, by applying Equation [9.1e], the phase φ could be obtained as:

$$\varphi = \tan^{-1}(\frac{y}{x}), \text{ if } x > 0 \text{ [Eq. 9.1g]}$$

$$\varphi = 2\pi + \tan^{-1}(\frac{y}{x}), \text{ if } x < 0 \text{ [Eq. 9.1h]}$$

2.4. Application fields

Polar coordinates are two-dimensional and thus they can be used only where point positions lie on a single two-dimensional plane (Batschelet, 1981). They are most appropriate in any context where the phenomenon being considered is inherently tied to direction and length from a center point. For instance: a) in defining mathematical curves (see Figure 9.1 above); b) in studying many physical systems (i.e. those concerned with bodies moving around a central point; Batschelet, 1975); c) in investigating the circular and orbital motion (i.e. in navigation, as the destination or direction of travel can be given as an angle and distance from the object being considered); d) in studying systems displaying radials symmetry that provide natural settings for the polar coordinate system, with the central point acting as the pole (i.e. the groundwater flow equation; systems with a radial force such as antenna); and e) in some industrials contexts.

2.5. A polar coordinate system to focus on the movement of the body centre of mass

As previously widely described in chapters 6 and 7, the three-dimensional displacement of the BCOM could be described by using suitable continuous functions such as Fourier Series.

Amplitude (A) and phase (φ) coefficients until the sixth harmonic (in each movement direction) best fit the 3D trajectories. Therefore, an average point (at a specific speed) of the displacement of the BCOM, could be defined as:

$$(A; \varphi \pm n \bullet 360°) \text{ [Eq. 9.2a]}$$

$$(-A; \varphi \pm (2n+1) \bullet 180°) \text{ [Eq. 9.2b]}$$

where *A* is the radial coordinate (or the distance, corresponding to the term *r* in some forms of the general equation = linear variable); *φ* is the angular anticlockwise coordinate (or the angle = angular variable); and *n* is any integer (Batschelet, 1975; 1981).

In our present case: **a)** the radial coordinate A corresponds to the amplitude value in the Fourier Series; and **b)** the angular coordinate φ to the phase value in the Fourier Series. As previously shown in chapter 7, all these values are known. As a result, along each movement direction, all average points (simultaneously, amplitude and phase coefficients) could be graphically represented in a polar logarithm graph.

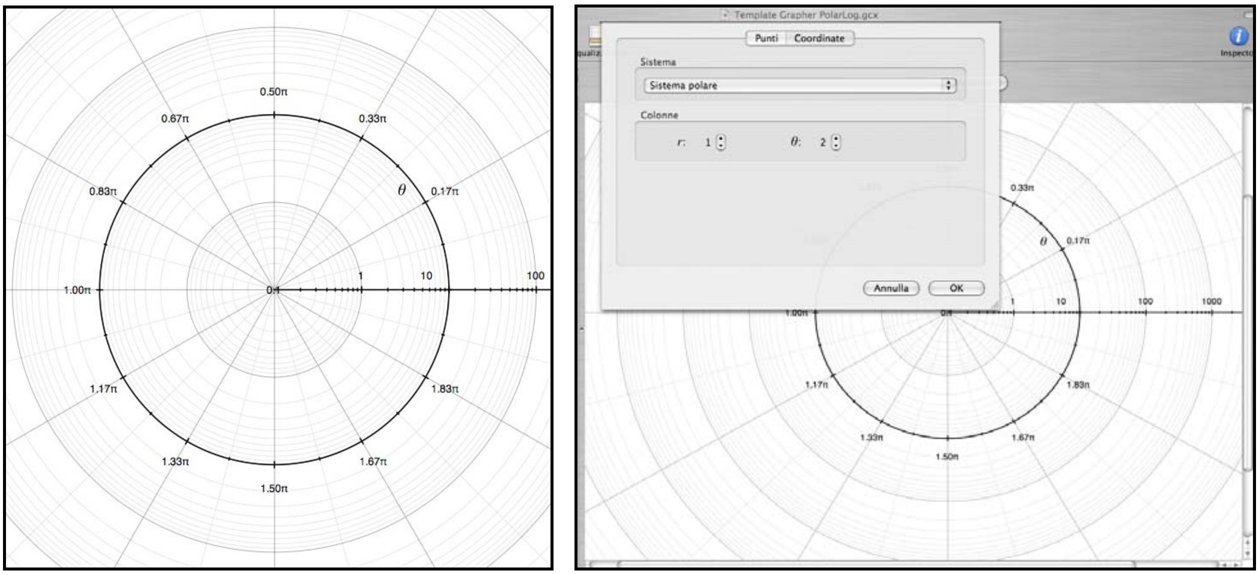
In the following paragraph, the main steps we followed to draw up a polar graph will be deeply illustrated and explained. Finally, we will focus on our results obtained in the different conditions.

3. POLAR LOGARITHM GRAPHS IN OUR STUDY

3.1. Graphical representation

Polar logarithm graphs were drawn up by the application *Grapher*, on a Macintosh notebook (by selecting the modality *log-polar system*; Figure 9.5). The circumference dimensions (0; 2π) were then selected (Batschelet, 1981); furthermore, the origin was $\log^{-1}$ (= 0.1).

a) Limits of the radial coordinate (distance) were: 0.1; 1 (first division); 10 (second division); and 100 (third division). These values of distances A agree with average values of harmonic coefficients obtained by Fourier analysis.
b) Intervals of the angular coordinate (angle) were π/6 (= 0.5235). In this way, the circumference was divided into 12 equal-dimensions segments.



Sistema: Sistema polare

Colonne: r: 1 θ: 2

Figure 9.5. A polar logarithm graph (on the left) and how data could be plotted into it (on the right).

3.2. Symmetrical and asymmetrical coefficients

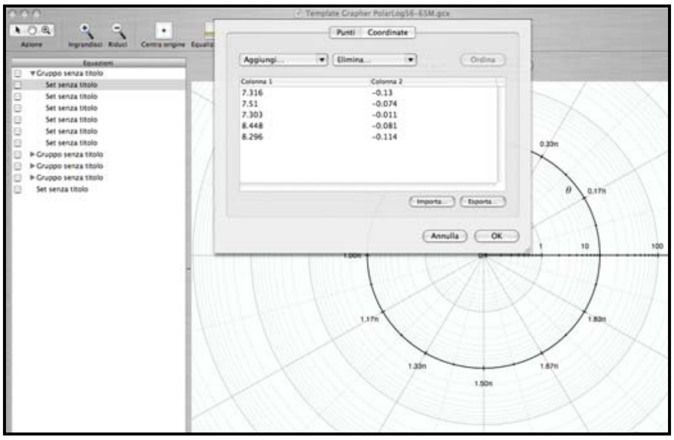
As briefly shown and discussed, we have decided to put into a polar graph only the first four harmonic coefficients (see also chapter 6, par. 5.2.2 and 7, par. 2.3). This decision is dependent on the fact the 5th and 6th harmonic are very small and their effect does not significantly influence on the pattern of the other coefficients. In this graphical analysis, we have therefore rejected the last two harmonics and their contribution.

Clearly, it is necessary to define a separate polar coordinate system along each movement direction (anterior/posterior, vertical and medial/lateral). We have then plotted in a polar logarithm graph both symmetrical and asymmetrical coefficients (amplitudes and phases). Mean values of these coefficients for each testing condition (as gender, age, type of gait, speed and gradient change) were used. To be more specific:

1. **symmetrical coefficients** ((A2; φ2) and (A4; φ4), along the anterior/posterior and vertical directions; (A1; φ1) and (A3; φ3), along the medial/lateral direction) were graphically plotted in each movement direction as speed increases, both in walking and running.

Anterior/posterior direction has been represented in blue (circles and line); vertical direction in orange (circles and line); and medial/lateral direction in grey (circles and line).

To be more precise, each circle corresponds to the mean value of the combination of the harmonic coefficients (amplitude A and phase φ; Figure 9.6) at a single speed; the line corresponds to a simple graphical union of these points, according to the increased movement speed; and the arrow point to the pattern direction of the coefficients.



Colonna 1	Colonna 2
7.316	-0.13
7.51	-0.074
7.303	-0.011
8.448	-0.081
8.296	-0.114

Figure 9.6. The combination of amplitudes and phases in a polar graph.

A graphical representation of the first symmetrical coefficients ((Ax2; φx2), (Ay2; φy2), and Az1; φz1)) has been presented in Figure 9.7; extreme velocities have been presented, as well.

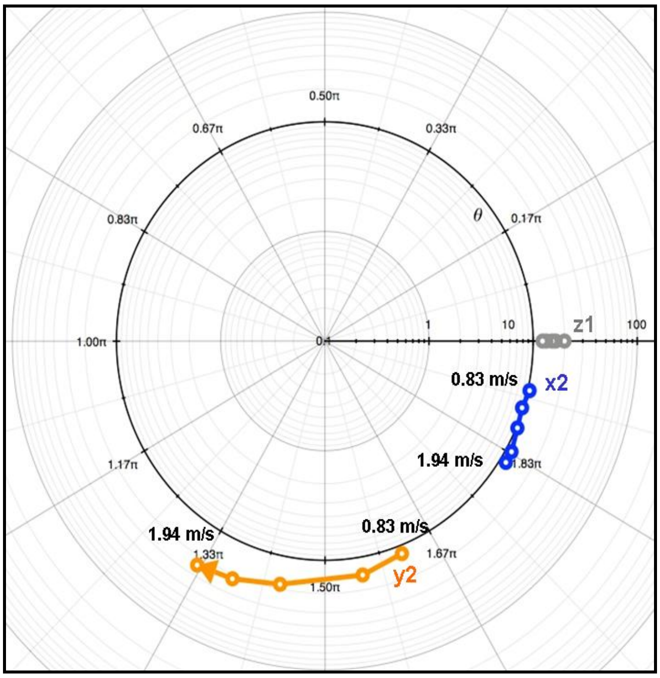


Figure 9.7. An example of symmetrical harmonic amplitudes and phases plotted in a polar graph.

2. **asymmetrical coefficients** (amplitudes and phases) for each movement direction ((A1; φ1) and (A3; φ3), along the anterior/posterior and vertical directions; (A2; φ2) and (A4; φ4) along the medial/lateral direction) were graphically plotted for each movement direction as speed increases, both in walking and running.

The graph legend is the same as already illustrated in symmetrical coefficients.

Consequently, polar-logarithm graphs have been drawn up in each testing condition.

3.3. Standard deviation of amplitude and phase coefficients

3.3.1. Amplitude standard deviation

Amplitude standard deviation is directly obtained from the Application called *Lissajous-Fourier BCOM Trajectory* (see also chapter 6, par. 2.1). More precisely, the knowledge of mean values of amplitude coefficients help in deriving the corresponding standard deviation values.

This is primarily due to the fact that the amplitude coefficient is a linear variable: indeed, it represents a distance (expressed in mm).

3.3.2. Phase standard deviation

Otherwise, it is more difficult to solve this problem in the case of phase. This is primarily due to the fact that the phase is a circular variable. In fact, it represents an angle: = arctan.2·(cosφ; sinφ) (see also chapter 6, par. 2.1).

Measures of concentration/dispersion (or straightness) of circular variables were proposed and discussed in Batschelet (1981). To define the standard deviation of the phase, we have to refer to the length of the mean vector *r* (e.g. in unimodal samples, the mean vector length *r* serves as a measure of concentration):

$$r = \sqrt{(\sin\varphi)^2 + (\cos\varphi)^2} \quad \text{or} \quad r = \frac{1}{n}[(\Sigma\cos\delta_i)^2 + (\Sigma\sin\delta_i)^2]^{\frac{1}{2}} \quad \text{[Eq. 9.3]}$$

where δ_i is derived by the difference among two angles.

The case of a maximum concentration occurs if the length of the mean vector *r* equals 1 (graph on the left top in Figure 9.8). Less concentration of the points leads to smaller values of *r*. The case of a minimum concentration occurs if the length of the mean vector *r* equals 0: no concentration around a single direction (graph on the right bottom in Figure 9.8; Batschelet, 1975).

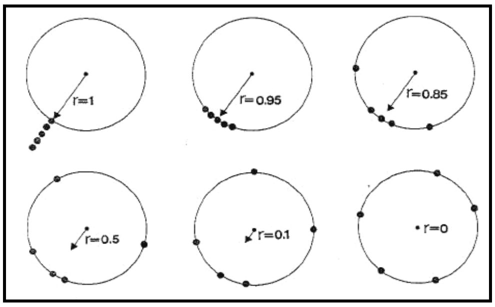


Figure 9.8. The range of the mean vector length, in Batschelet (1981).

Moreover, as also suggested in Batschelet (1981), the mean vector length *r* could be used as a measure of concentration. Indirectly, this is also a measure of dispersion. Since *r* decreases from 0 to 1 while the dispersion increases, it seems to be natural to consider *1-r* as a measure of dispersion (Batschelet, 1981). However, we should consider the term *2·(1-r)* rather than *1-r* as a suitable statistical tool. Therefore, the quantity:

$$s^2 = \frac{1}{n} \bullet \Sigma 2 \bullet [1-(\cos\varphi_1 - \bar{\varphi})] \text{ [Eq. 9.4a] or } s^2 = 2 \bullet (1-r) \text{ [Eq. 9.4b]}$$

constitutes the *angular variance* (i.e. asymptotically equivalent to the variance in linear statistics; Batschelet, 1965). Taking the square root (*mean angular deviation* or *angular deviation*), a measure of dispersion is obtained:

$$s = \frac{1}{n} \bullet \Sigma 2 \bullet [1-(\cos\varphi_1 - \bar{\varphi})]^{1/2} \text{ [Eq. 9.4c] or}$$

$$s = [2 \bullet (1-r)]^{1/2} \text{ [Eq. 9.4d] and } s = \frac{180°}{\pi} \bullet [2 \bullet (1-r)]^{1/2} \text{ [Eq. 9.4e]}$$

To be precise, if *r* equals 0, *s* is √2 (≈ 1.41); however, if *r* equals 1, *s* becomes 0. Therefore, *s* ranges from a minimum of 0 to a maximum of ≈ 1.41.

Both angular variance and mean angular deviation are equivalent to the corresponding measures in linear statistics. As thought and demonstrated, the angular deviation increases as the mean vector decreases (Figure 9.9).

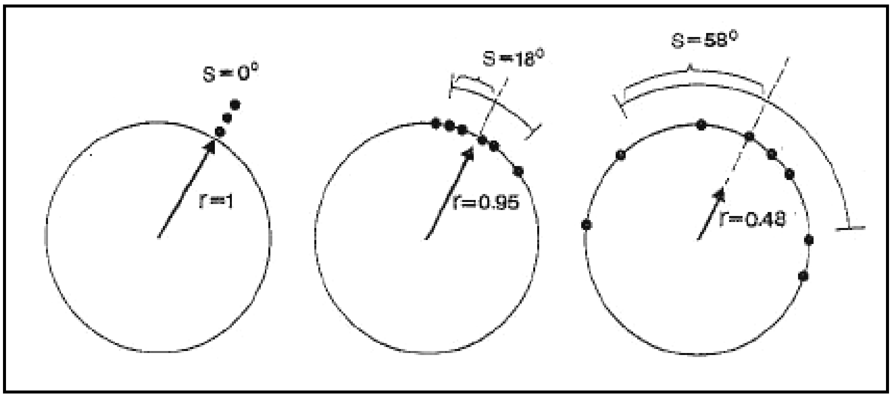


Figure 9.9. The relationship between mean vector length and angular deviation, in Batschelet (1981).

To sum up, we have calculated the standard deviation of the phase by applying Equation [9.4e].

3.3.3. Graphical representation of standard deviations

In order to represent standard deviations of both amplitude (A) and phase (φ) coefficients, the mean values of those have to be considered. Moreover, the corresponding mean values of standard deviations (see par. 3.3.1 and 3.3.2 above) have to take into account.

The knowledge of these parameters allows us to define 5 new points that summarize all their information (mean ± S.D.):

point 1 = ((A + S.D.A); φ)
point 2 = (A; (φ + S.D.φ))
point 3 = ((A - S.D.A); φ)
point 4 = (A; (φ - S.D.φ))
point 5 = ((A + S.D.A); φ)

Thus, the graphical grouping of these points constitutes the confidence interval of each mean value (at a single speed; Batschelet, 1981). This grouping looks like a quadrilateral.

In Figure 9.10, an example of this graphical confidence interval has been proposed for Ax2 and φx2 (Figure a); Ay2 and φy2 (Figure b); finally, Az3 and φz3 (Figure c). We have decided to take this last symmetrical coefficient (Az3; φz3) because, clearly, the standard deviation of the other symmetrical coefficient (Az1; φz1) equals 0. This is due to the fact that the mean vector *r* equals 1.

These graphical representations have been proposed in all males aged 25 to 35 who walk on the level ground at the speed of 0.83 m/s.

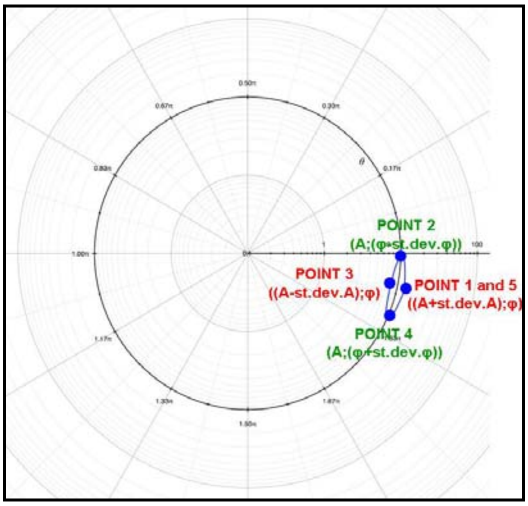


Figure 9.10a. Confidence interval of mean coefficient x2 (Ax2; φx2), in level walking at 0.83 m/s.

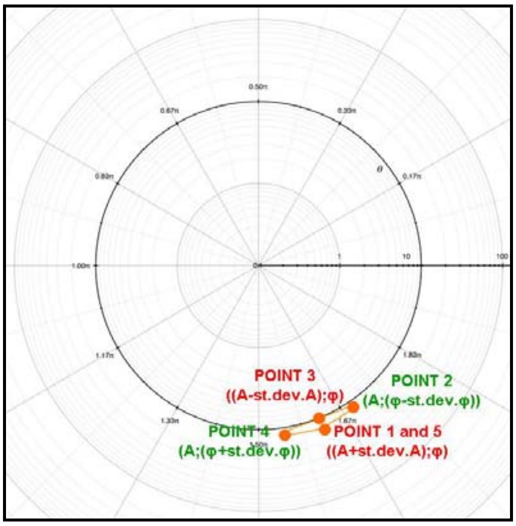


Figure 9.10b. Confidence interval of mean coefficient y2 (Ay2; φy2), in level walking at 0.83 m/s.

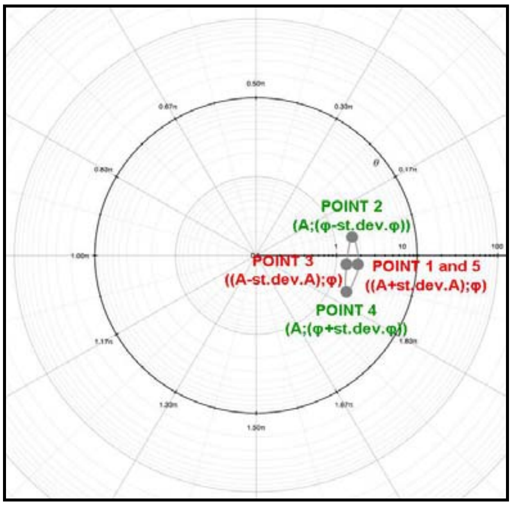


Figure 9.10c. Confidence interval of mean coefficient z3 (Az3; φz3), in level walking at 0.83 m/s.

According to what previously discussed in chapter 7, these graphical representations were just made for the main three age groups: 1) subjects (males and females) aged 6 to 13; 2) subjects aged 25 to 35; and 3) subjects aged 56 to 65.

Results of the statistical analysis have been already presented and discussed in Appendix 7.1.

Clearly, it would be possible to extend these information to the other age groups and gait gradient conditions, too. In detail, all radii and standard deviations (in all age groups, males and females) are contained in the enclosed CD (First Study, Chapter 9, Radii variable: variance and deviation; coefficients mean, standard deviation and coefficients range polar log graphs).

4. RESULTS OF OUR STUDY

4.1. Fourier coefficients in a polar graph as a function of gender

Only a qualitative approach has been applied upon our polar graphs.

In **level walking**, our results show that, on average, independently of both movement direction and age, there are no significant differences in all the symmetrical coefficients (x2, x4, y2, y4 and z1, z3) between males and females. Because of the absence of evident differences among gender, no graphical examples have been proposed.

As far as asymmetrical coefficients have been concerned, our results show that on average (in all age groups): a) in forward direction, asymmetrical coefficient x1 seems to be downward right shifted in males than in females; b) in vertical direction, y1 seems to be slightly downward left shifted in males than in females; and c) in lateral direction, z2 seems to be left shifted in males than in females.

A comprehensive graphical example is illustrated in Figure 9.11: only average points in males aged 25 to 35, at all speeds (left graph) and in females aged 25 to 35 (right graph) have been drawn. A similar pattern occurs in the other age groups.

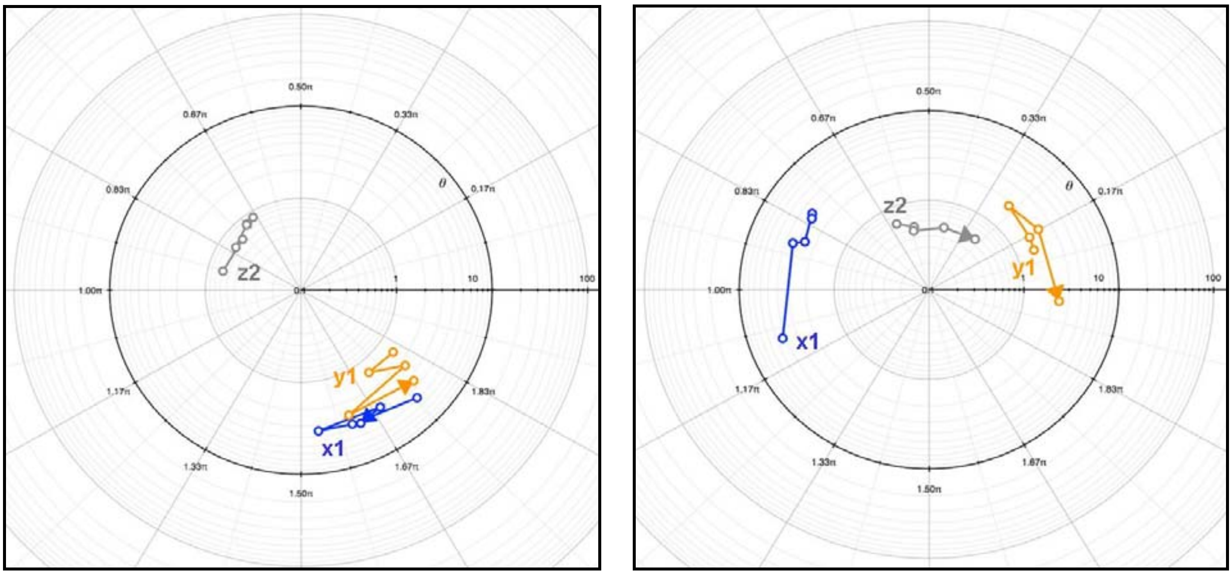


Figure 9.11. Average points in x1, y1 and z2 in level walking at all speeds, males (on the left) and females (on the right) aged 25 to 35.

Moreover, independently of both movement direction and age, there are no significant differences in the other asymmetrical coefficients (x3, y3 and z4) between males and females. Because of the absence of evident differences among gender, no graphical examples have been proposed.

In **level running**, as far as all symmetrical coefficients have been concerned, our results show that on average (in all age groups): a) in forward and lateral directions, symmetrical coefficients x2, x4 and z1, z3 seem to have a slightly wider range (in the same graphical section) in males than in females; however, b) in vertical direction, y2 and y4 seem to have a slightly wider range (in the same graphical section) in females than in males.

A comprehensive graphical example is illustrated in Figure 9.12: only average points in males aged 25 to 35, at all speeds (left graph) and in females aged 25 to 35 (right graph) have been drawn. A similar pattern occurs in the other age groups.

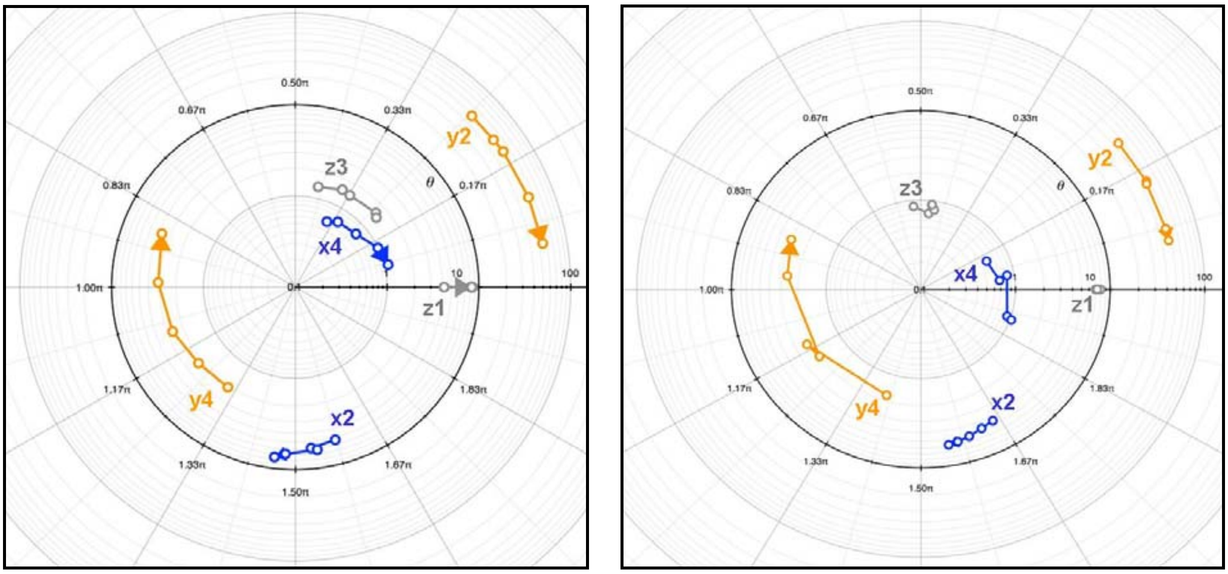


Figure 9.12. Average points in x2, x4, y2, y4 and z1, z3 in level running at all speeds, males (on the left) and females (on the right) aged 25 to 35.

Moreover, as far as all <u>asymmetrical coefficients</u> have been concerned, our results show that on average (in all age groups): a) in forward direction, asymmetrical coefficient x1 seems to be slightly downward shifted in males than in females; b) in vertical direction, y1 seems to be downward left shifted in males than in females; and c) in lateral direction, z2 seems to slightly downward left shifted in males than in females.

A comprehensive graphical example is illustrated in Figure 9.13: only average points in males aged 25 to 35, at all speeds (left graph) and in females aged 25 to 35 (right graph) have been drawn. A similar pattern occurs in the other age groups.

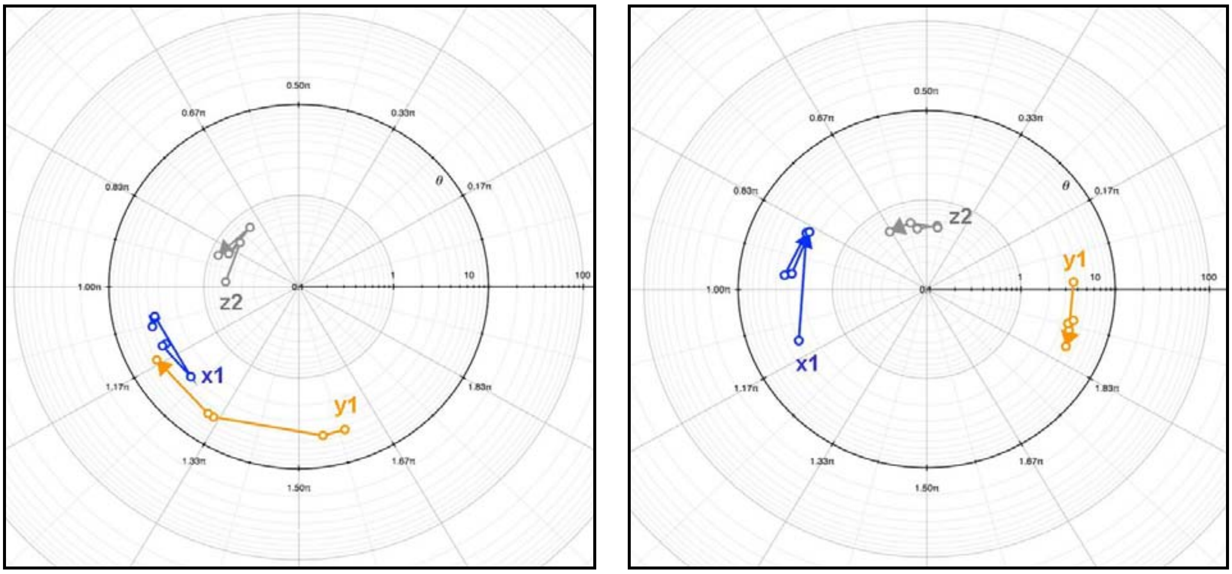


Figure 9.13. Average points in x1, y1 and z2 in level running at all speeds, males (on the left) and females (on the right) aged 25 to 35.

Moreover, independently of both movement direction and age, there are no significant differences in the other asymmetrical coefficients (x3, y3 and z4) between males and females. Because of the absence of evident differences among gender, no graphical examples have been proposed.

In conclusion, only limited differences could be distinguished between males and females independently of harmonic coefficient, gait, age and movement direction.

Both in gradient walking and running, males and females seem not to differ in a significant way. For more information, see par. 4.4 below.

4.2. Fourier coefficients in a polar graph as a function of age

A qualitative approach has been applied upon our polar graphs, as well.

In the following tables, the main peculiarities along each movement direction (walking and running) are summarized as a function of age in males (Table a) and females (Table b), respectively. More precisely, general details referred to the range of a single coefficient are illustrated and presented.

We have decided to consider the middle class (subjects aged 25 to 35) as a reference point for many reasons: a) independently of testing conditions, there are only little differences among gender (see also par. 4.1 above); b) in comparison to the other age groups, the pattern of this age group is the most clear, definite and regular (see also chapter 7); and c) it seems that the middle class better reflects the behaviour of the optimal (and expected) condition and of the other ages.

Consequently, details and observations reported in all tables are related to this age group (males and females). In addition: a) the writing ‘very similar’ means that there are no relevant differences in both range limits and contours while ‘quite similar’ means that there are only little differences; b) the writing ‘more confined range’ means that the examined range is limited compared to subjects aged 25 to 35; c) the writing ‘anomalous pattern’ means that there are only little similarities by comparing the two graphs; d) the writing ‘upward shift’ means that all coefficients have been moved upward along the circumference while ‘downward shift’ corresponds to the movement in the opposite direction; finally, e) the writing ‘shifted in the opposite section’ means that the graphs have been moved into the mirrored quadrant; and so on.

Therefore, the tables and the reference polar graphs have been reported below. Otherwise, it is important to remember that the most complete information could be obtained by only deeply looking at single polar graphs.

In **level walking**:

A. Herein there are the reference graphs regarding the symmetrical coefficients (x2, y2 and z1) in males (left graph) and females (right graph) aged 25 to 35:

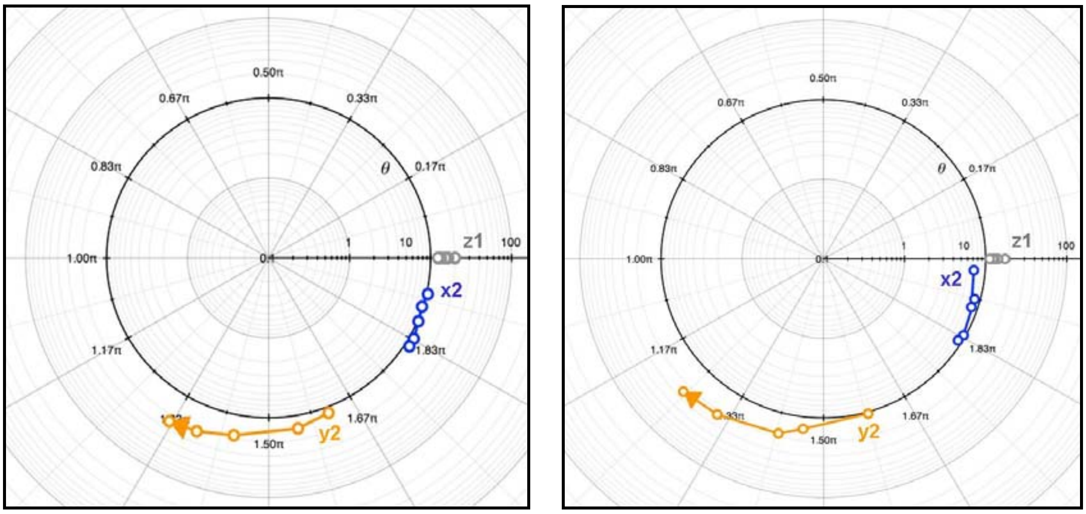


Figure 9.14. Average points in x2, y2 and z1 in level walking,
males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

SYMMETRICAL COEFFICIENTS	**6 - 13 y (MALES)**	**14 - 17 y**	**18 - 24 y**
x2 (forward)	more confined range	more confined range (similar to 6 - 13 y)	upward shift (0π; 0.17π)
y2 (vertical)	very similar	very similar	upward shift (1.83π; 1.33π)
z1 (lateral)	more confined range	very similar	very similar
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x2	upward shift (0π; 1.83π)	more confined range (similar to 6 - 13 y)	starts from point 0
y2	very similar	very similar	very similar
z1	very similar	more confined range (1.67π; 1.50π)	very similar

Table 9.1a. x2, y2 and z1 pattern in level walking, all age groups (males).

SYMMETRICAL COEFFICIENTS	**6 - 13 y (FEMALES)**	**14 - 17 y**	**18 - 24 y**
x2 (forward)	very similar	very similar	very similar
y2 (vertical)	very similar	very similar	very similar
z1 (lateral)	slightly wider range	very similar	slightly wider range
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x2	slightly upward shift	very similar	very similar
y2	very similar	very similar	very similar
z1	very similar	very similar	very similar

Table 9.1b. x2, y2 and z1 pattern in level walking, all age groups (females).

B. In addition, herein there are the reference graphs regarding the other symmetrical coefficients (x4, y4 and z3) in males (left graph) and females (right graph) aged 25 to 35:

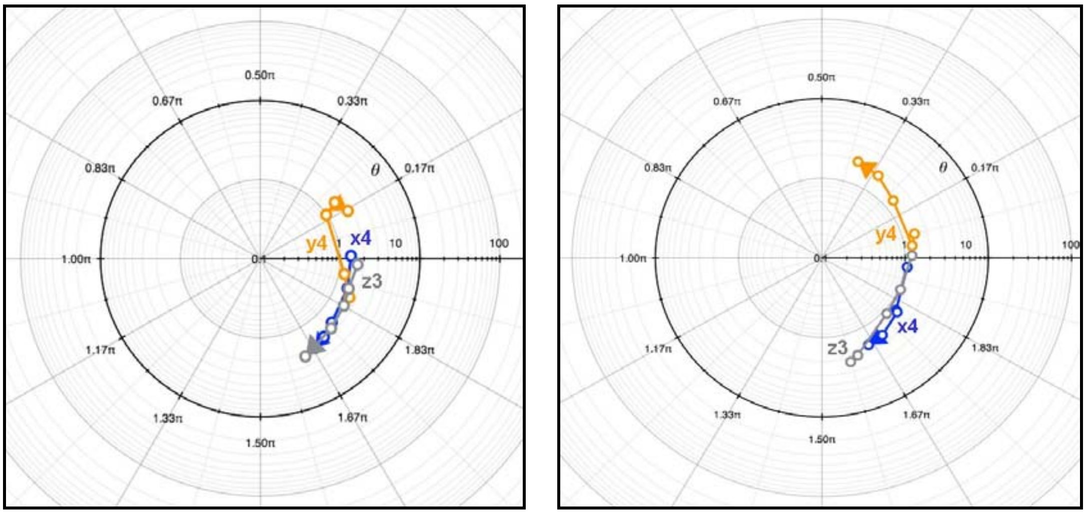


Figure 9.15. Average points in x4, y4 and z3 in level walking, males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

SYMMETRICAL COEFFICIENTS	6 - 13 y (MALES)	14 - 17 y	18 - 24 y
x4 (forward)	slightly upward shift (0.17π; 1.83π)	slightly upward shift (0.33π; 1.83π)	more confined range (0.33π; 0.17π)
y4 (vertical)	slightly wider range	slightly more confined range	upward shift (0.67π; 0.33π)
z3 (lateral)	very similar	slightly wider range	slightly wider range
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x4	upward shift (0.17π; 1.83π)	more confined range (0.17π; 0π)	upward shift (0.17π; 1.83π)
y4	slightly wider range	wider range (starts from 1.83π)	very similar
z3	very similar	very similar	very similar

Table 9.2a. x4, y4 and z3 pattern in level walking, all age groups (males).

SYMMETRICAL COEFFICIENTS	6 - 13 y (FEMALES)	14 - 17 y	18 - 24 y
x4 (forward)	very similar	very similar	wider range (stops to 1.50π)
y4 (vertical)	shifted to the opposite section	downward shift (0.17π; 1.83π)	data approaching 0π
z3 (lateral)	very similar	very similar	quite similar
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x4	data approaching 0π	very similar	very similar
y4	very similar	wider range (starts from 1.83π)	very similar
z3	very similar	slightly more confined range	more confined range

Table 9.2b. x4, y4 and z3 pattern in level walking, all age groups (females).

C. Furthermore, herein there are the reference graphs in the asymmetrical coefficients (x1, y1 and z2) in males (left graph) and females (right graph) aged 25 to 35:

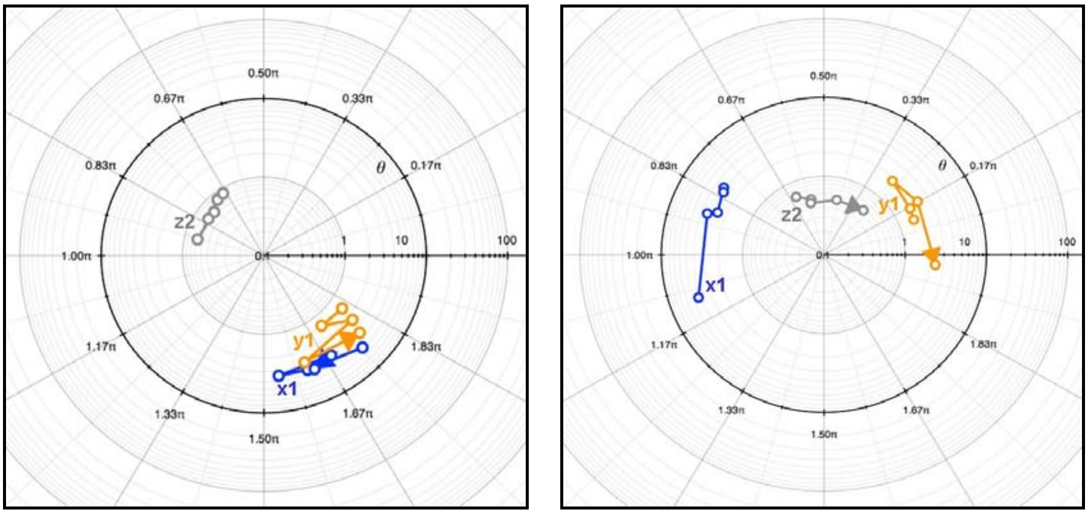


Figure 9.16. Average points in x1, y1 and z2 in level walking,

males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

ASYMMETRICAL COEFFICIENTS	**6 - 13 y (MALES)**	**14 - 17 y**	**18 - 24 y**
x1 (forward)	shifted to the upward section (0.33π; 1.83π)	shifted to the upward section (0.17π; 1.83π)	anomalous pattern
y1 (vertical)	wider range (0.33π; 1.83π)	anomalous pattern	anomalous pattern
z2 (lateral)	anomalous pattern	slightly wider range	right shift (0.67π; 0.17π)
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x1	shifted to the opposite section (1.00π; 0.50π)	wider upward range (0.17π; 1.67π)	very similar
y1	shifted to the opposite section (0.83π; 0.17π)	shifted to the upward section (0.50π; 0.17π)	anomalous pattern
z2	very similar	very similar (data approaching 0.50π)	very similar

Table 9.3a. x1, y1 and z2 pattern in level walking, all age groups (males).

ASYMMETRICAL COEFFICIENTS	**6 - 13 y (FEMALES)**	**14 - 17 y**	**18 - 24 y**
x1 (forward)	anomalous pattern	shifted to the opposite section	very similar (data approaching 1.33π)
y1 (vertical)	slightly wider range	shifted to the opposite section (1.33π; 1.00π)	very similar
z2 (lateral)	slightly wider range	wider range (1.17π; 0.50π)	quite similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x1	anomalous pattern	upward shift (0.83π; 0.50π)	shifted to the opposite section
y1	wider range (1.17π; 0.67π)	anomalous pattern	quite similar
z2	quite similar	left downward shift (1.50π; 0.83π)	very similar

Table 9.3b. x1, y1 and z2 pattern in level walking, all age groups (females).

D. In addition, herein there are the reference graphs in the other asymmetrical coefficients (x3, y3 and z4) in males (left graph) and females (right graph) aged 25 to 35:

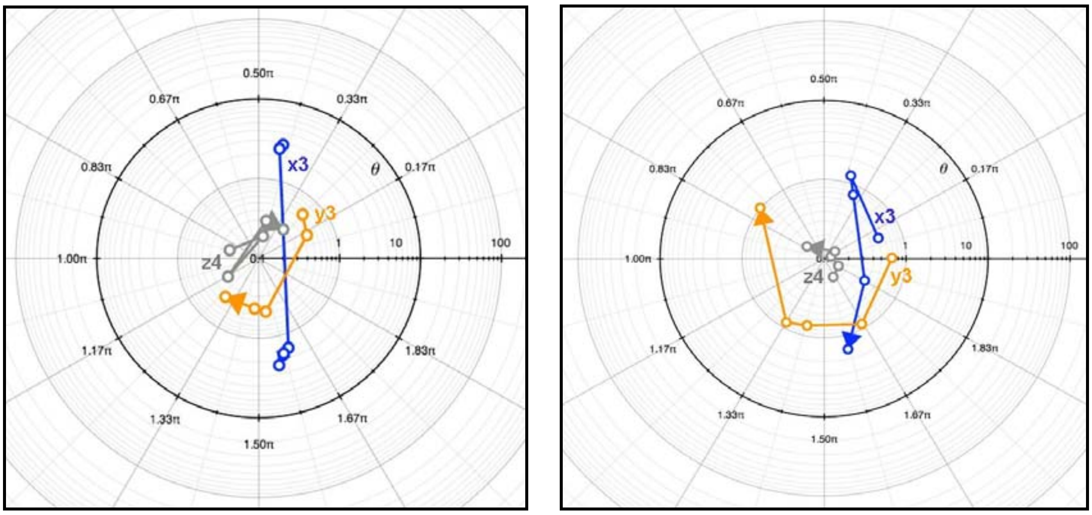


Figure 9.17. Average points in x3, y3 and z4 in level walking,
males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

ASYMMETRICAL COEFFICIENTS	**6 - 13 y (MALES)**	**14 - 17 y**	**18 - 24 y**
x3 (forward)	anomalous pattern	quite similar	slightly more confined range
y3 (vertical)	more confined range (0.17π; 0.67π)	shifted to the opposite section	very similar
z4 (lateral)	quite similar	very similar	very similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x3	quite similar	quite similar	more confined range
y3	left upward shift (0.67π; 1.50π)	shifted to the opposite section	more confined range (1.33π; 1.17π)
z4	very similar	more confined range (1.50π; 1.67π)	anomalous pattern (data approaching 0π)

Table 9.4a. x3, y3 and z4 pattern in level walking, all age groups (males).

ASYMMETRICAL COEFFICIENTS	**6 - 13 y (FEMALES)**	**14 - 17 y**	**18 - 24 y**
x3 (forward)	anomalous pattern	anomalous pattern	quite similar (data approaching 1.30π)
y3 (vertical)	slightly shift to the upward section	quite similar	quite similar
z4 (lateral)	quite similar	very similar	very similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x3	anomalous pattern	quite similar	quite similar
y3	quite similar	quite similar	quite similar
z4	very similar	very similar	very similar

Table 9.4b. x3, y3 and z4 pattern in level walking, all age groups (females).

In **level running**:

A. Herein there are the reference graphs in the symmetrical coefficients (x2, y2 and z1) in males (left graph) and females (right graph) aged 25 to 35:

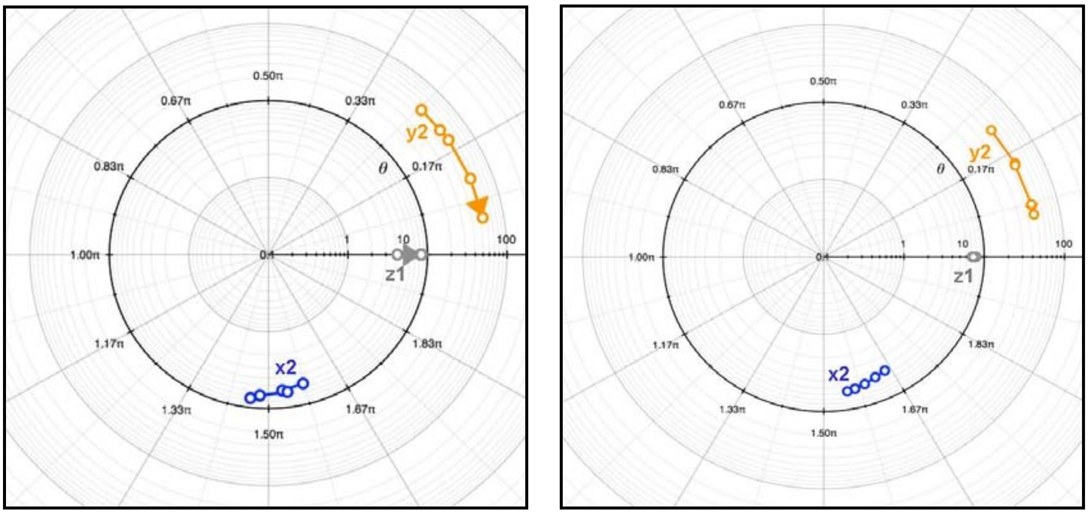


Figure 9.18. Average points in x2, y2 and z1 in level running, males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

SYMMETRICAL COEFFICIENTS	6 - 13 y (MALES)	14 - 17 y	18 - 24 y
x2 (forward)	downward shift (1.83π; 1.67π)	downward shift (1.50π; 1.17π)	anomalous pattern
y2 (vertical)	downward shift (0.50π; 0.17π)	very similar	very similar
z1 (lateral)	very similar	very similar	anomalous pattern
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x2	very similar	downward shift (1.83π; 1.67π)	downward shift (1.83π; 1.67π)
y2	slightly wider range	upward shift (1.67π; 1.33π)	downward shift (1.50π; 1.17π)
z1	very similar	very similar	very similar

Table 9.5a. x2, y2 and z1 pattern in level running, all age groups (males).

SYMMETRICAL COEFFICIENTS	6 - 13 y (FEMALES)	14 - 17 y	18 - 24 y
x2 (forward)	downward shift (1.83π; 1.67π)	slightly wider range	slightly wider range (similar to 14 - 17 y)
y2 (vertical)	more confined range	wider range	wider range
z1 (lateral)	slightly downward shift	very similar	very similar
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x2	downward shift (1.83π; 1.67π)	slightly downward shift	very similar
y2	slightly wider range	slightly downward shift	slightly downward shift
z1	very similar	very similar	very similar

Table 9.5b. x2, y2 and z1 pattern in level running, all age groups (females).

B. In addition, herein there are the reference graphs in the other symmetrical coefficients (x4, y4 and z3) in males (left graph) and females (right graph) aged 25 to 35:

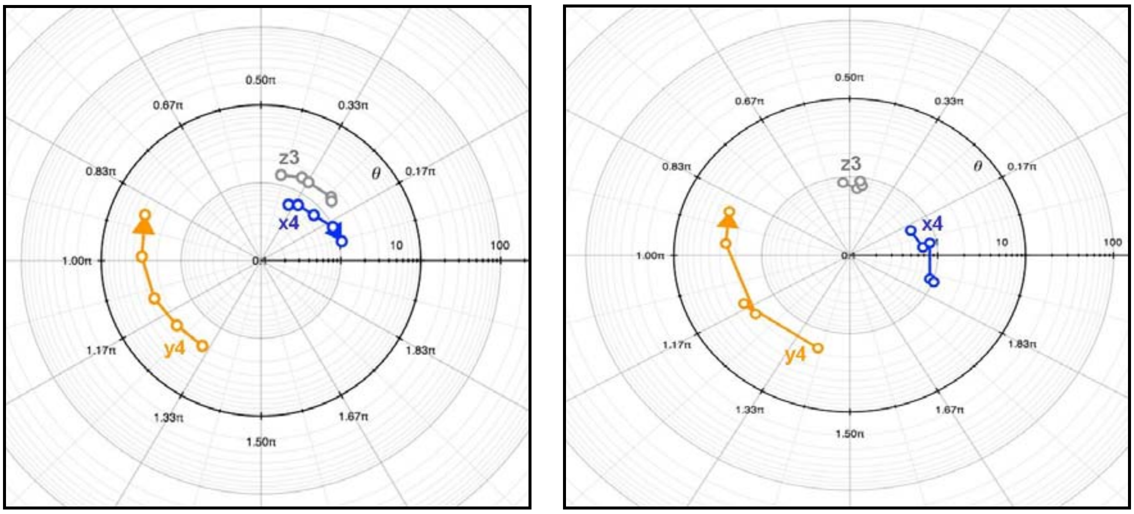


Figure 9.19. Average points in x4, y4 and z3 in level running, males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

SYMMETRICAL COEFFICIENTS	**6 - 13 y (MALES)**	**14 - 17 y**	**18 - 24 y**
x4 (forward)	upward left shift (0.67π; 0.17π)	wider range (0.33π; 1.67π)	very similar
y4 (vertical)	downward shift (1.50π; 1.17π)	wider range (1.33π; 0.50π)	very similar
z3 (lateral)	wider range (0.83π; 0.17π)	very similar	very similar
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x4	very similar	upward left shift (0.83π; 0.33π)	very similar (data more close)
y4	wider range (stops to 0.67π)	upward left shift (1.67π; 1.17π)	very similar (data more close)
z3	upward left shift (0.67π; 0.33π)	very similar	very similar (data more close)

Table 9.6a. x4, y4 and z3 pattern in level running, all age groups (males).

SYMMETRICAL COEFFICIENTS	**6 - 13 y (FEMALES)**	**14 - 17 y**	**18 - 24 y**
x4 (forward)	upward left shift (0.50π; 0.17π)	very similar	downward shift (1.83π; 1.33π)
y4 (vertical)	more confined range (data approaching 1.33π)	shifted to the opposite section	downward shift (1.50π; 1.17π)
z3 (lateral)	wider range	very similar	very similar
SYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x4	very similar	quite similar	shifted to the opposite section
y4	shifted to the opposite section	upward right shift (0.17π; 1.67π)	upward right shift (1.83π; 1.67π)
z3	very similar	wider range (0.33π; 0.83π)	wider range (0.67π; 1.00π)

Table 9.6b. x4, y4 and z3 pattern in level running, all age groups (females).

C. Furthermore, herein there are the reference graphs in the asymmetrical coefficients (x1, y1 and z2) in males (left graph) and females (right graph) aged 25 to 35:

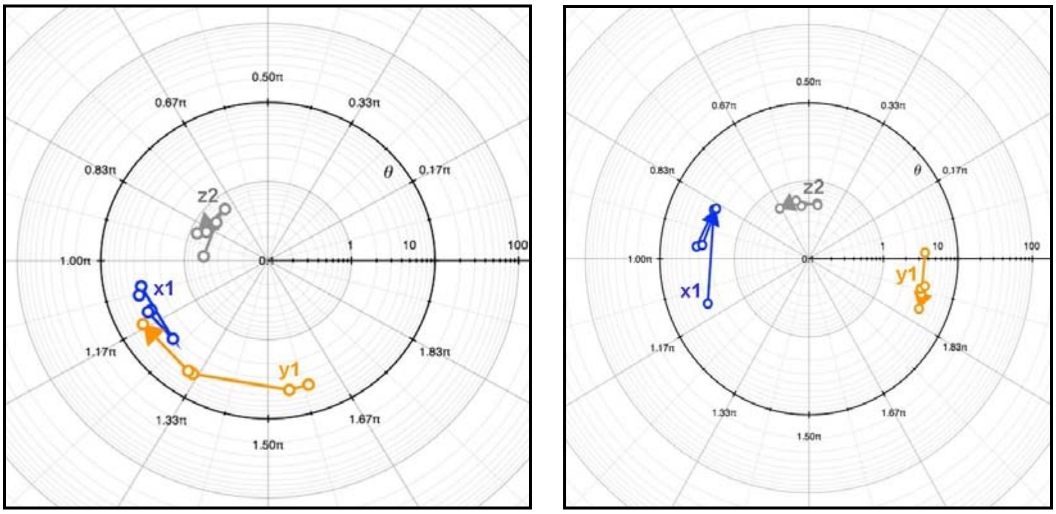


Figure 9.20. Average points in x1, y1 and z2 in level running, males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

ASYMMETRICAL	6 - 13 y (MALES)	14 - 17 y	18 - 24 y
x1 (forward)	wider range (1.50π; 1.33π)	wider range	wider range
y1 (vertical)	upward right shift (1.83π; 1.67π)	upward left shift (0.83π; 0.33π)	more confined range
z2 (lateral)	upward right shift (0.67π; 0.33π)	quite similar	anomalous pattern
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x1	very similar	shifted to the opposite section (0.33π; 1.83π)	downward shift (1.50π; 1.33π)
y1	very similar	upward left shift (1.83π; 0.17π)	more confined range
z2	anomalous pattern	anomalous pattern	shifted to the opposite section (0.17π; 1.67π)

Table 9.7a. x1, y1 and z2 pattern in level running, all age groups (males).

ASYMMETRICAL COEFFICIENTS	6 - 13 y (FEMALES)	14 - 17 y	18 - 24 y
x1 (forward)	anomalous pattern	downward shift (1.50π; 1.17π)	quite similar
y1 (vertical)	anomalous pattern	anomalous pattern	anomalous pattern
z2 (lateral)	anomalous pattern	quite similar (data approaching 0.83π)	very similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x1	anomalous pattern	upward right shift (0.83π; 0.33π)	downward right shift (1.83π; 1.50π)
y1	shifted to the opposite section	shifted to the opposite section (1.50π; 1.00π)	quite similar
z2	wider range	anomalous pattern	very similar

Table 9.7b. x1, y1 and z2 pattern in level running, all age groups (females).

D. In addition, herein there are the reference graphs in the other asymmetrical coefficients (x3, y3 and z4) in males (left graph) and females (right graph) aged 25 to 35:

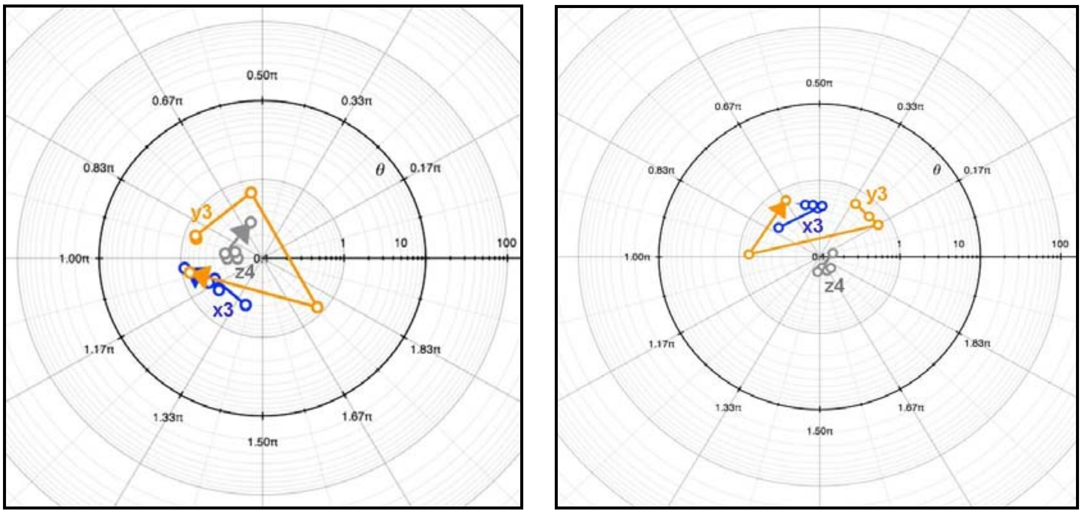


Figure 9.21. Average points in x3, y3 and z4 in level running, males (on the left) and females (on the right) aged 25 to 35.

Specifically, compared to this reference age, our results show that the other groups have:

ASYMMETRICAL COEFFICIENTS	6 - 13 y (MALES)	14 - 17 y	18 - 24 y
x3 (forward)	shifted to the opposite section (0.67π; 0.33π)	wider range (1.17π; 0.33π)	shifted to the opposite section (0.50π; 1.83π)
y3 (vertical)	anomalous pattern	quite similar	quite similar
z4 (lateral)	quite similar	downward right shift (1.50π; 1.00π)	very similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x3	shifted to the opposite section (0.67π; 0.33π)	data approaching 1.67π	shifted to the opposite section (data approaching 0.50π)
y3	very similar	data approaching 0.50 and 0.33π	anomalous pattern
z4	very similar	quite similar	very similar

Table 9.8a. x3, y3 and z4 pattern in level running, all age groups (males).

ASYMMETRICAL COEFFICIENTS	6 - 13 y (FEMALES)	14 - 17 y	18 - 24 y
x3 (forward)	anomalous pattern	quite similar	anomalous pattern
y3 (vertical)	quite similar	anomalous pattern	anomalous pattern
z4 (lateral)	very similar	very similar	very similar
ASYMMETRICAL	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
x3	anomalous pattern	quite similar	quite similar
y3	anomalous pattern	anomalous pattern	data approaching 0.17π
z4	very similar	quite similar	very similar

Table 9.8b. x3, y3 and z4 pattern in level running, all age groups (females).

4.3. Fourier coefficients in a polar graph as a function of speed

In all testing conditions, the factor ‘speed’ plays an important role in defining and characterizing the pattern of harmonic coefficients. However, it is quite difficult to describe its part only with words. In fact, a polar graphical example better summarizes all the peculiarities and properties of such a movement.

Thus, the following polar graphs visualize the important action of speed both in walking and running, in all movement directions. We have decided to present and discuss graphs (average points and corresponding standard deviations) referred to males aged 25 to 35 (see par. 4.2 above).

In **level walking**, symmetrical coefficients (x2, x4, y2, y4 and z1, z3) show that:

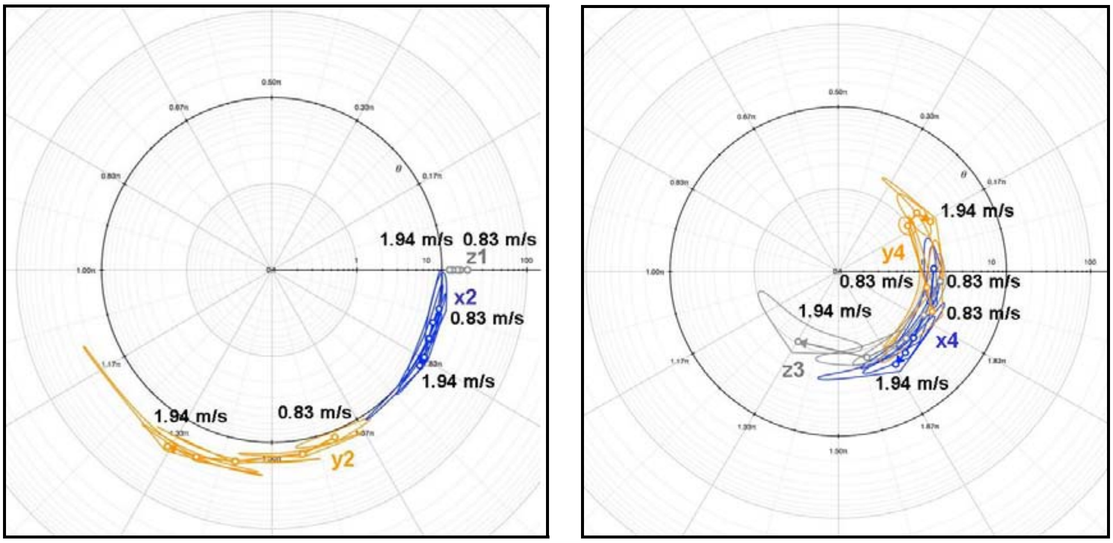


Figure 9.22. Average points and standard deviations in x2, y2 and z1 (on the left) and in x4, y4 and z3 (on the right) in level walking at all speeds, males aged 25 to 35.

- in x2, y2 and z1: the factor ‘speed’ produces a significant left shift. In other words, if walking speed increases, each symmetrical coefficient progressively moves towards left. Clearly, this pattern is less marked in lateral direction;
- moreover, in x4 and z3: the factor ‘speed’ produces a significant left shift. However, in y4 it produces a slightly right upward shift. In other words, if walking speed increases, this coefficient progressively moves above and towards right.

In addition, <u>asymmetrical coefficients</u> (x1, x3, y1, y3 and z2, z4) show that:

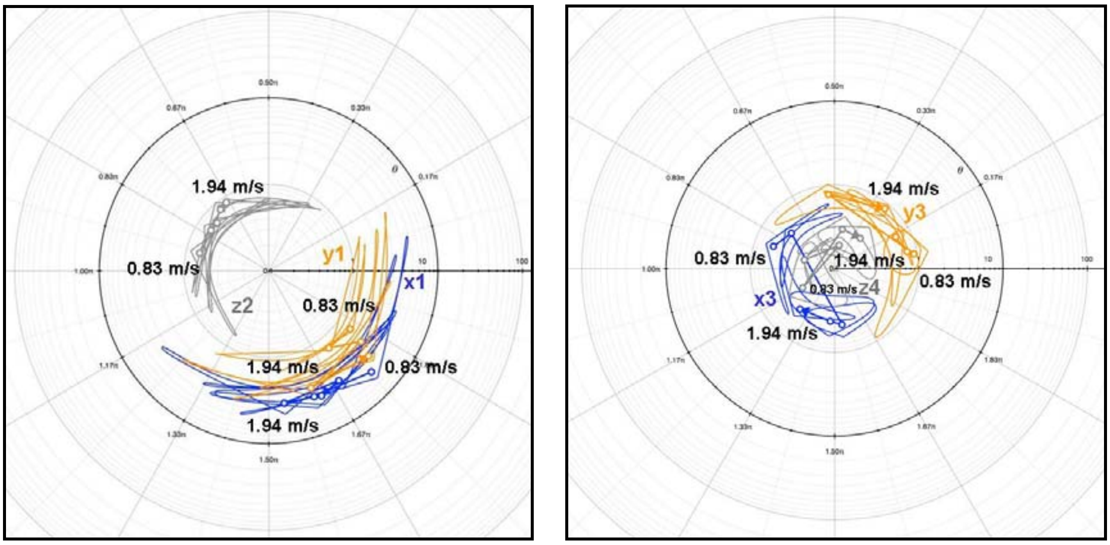


Figure 9.23. Average points and standard deviations in x1, y1 and z2 (on the left) and in x3, y3 and z4 (on the right) in level walking at all speeds, males aged 25 to 35.

- in x1, y1 and z2: the factor ‘speed’ produces a significant left shift;

- moreover, in x3 and y3: the factor ‘speed’ produces a significant right downward shift. In other words, if walking speed increases, each coefficient progressively moves below and towards right. However, in z4 the factor ‘speed’ produces a slightly right upward shift.

In **level running**, symmetrical coefficients (x2, x4, y2, y4 and z1, z3) show that:

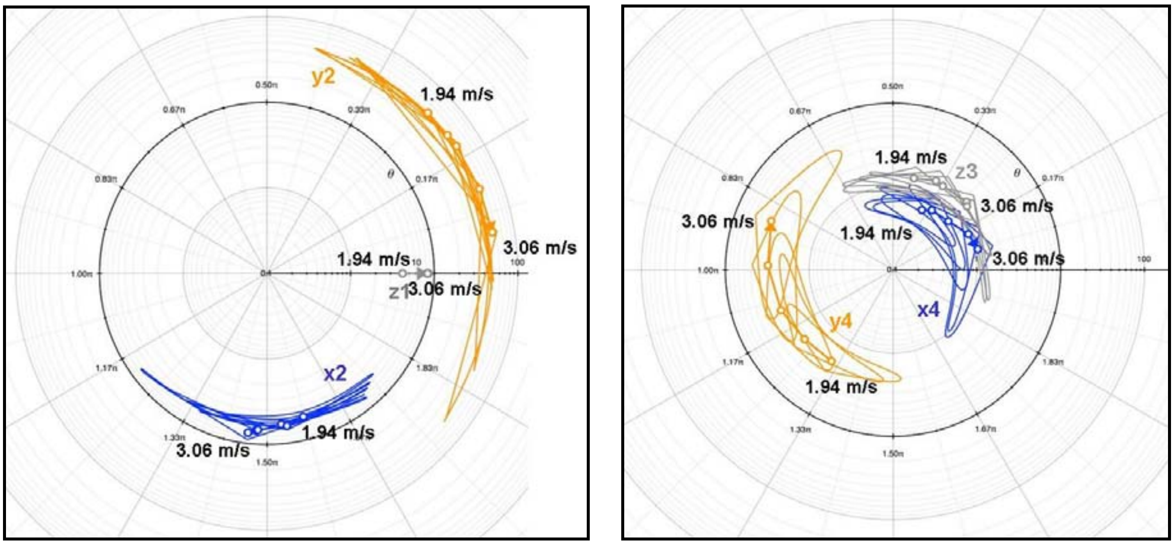


Figure 9.24. Average points and standard deviations in x2, y2 and z1 (on the left) and in x4, y4 and z3 (on the right) in level running at all speeds, males aged 25 to 35.

- in x2 and y2: the factor ‘speed’ produces a significant left shift. However, in z1 it produces a right shift;
- moreover, in x4, y4 and z3: the factor ‘speed’ produces a significant left shift.

In addition, asymmetrical coefficients (x1, x3, y1, y3 and z2, z4) show that:

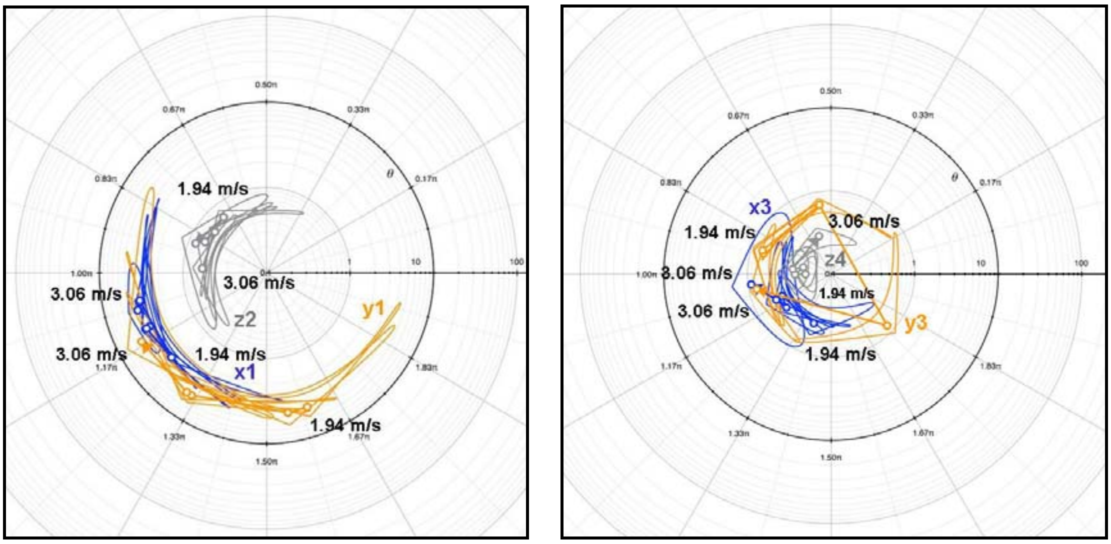


Figure 9.25. Average points and standard deviations in x1, y1 and z2 (on the left) and in x3, y3 and z4 (on the right) in level running at all speeds, males aged 25 to 35.

- in x1 and y1: the factor 'speed' produces a significant left shift. However, in z2 it produces a right downward shift;
- moreover, in x3 and z4: the factor 'speed' produces a significant left shift. However, in y3 it produces a right downward shift.

In conclusion, independently of testing condition (harmonic coefficient, gender, age and gait), the factor 'speed' plays the most important role (and action) in characterizing the movement (or shift) along the circumference.

4.4. Fourier coefficients in a polar graph as function of gradient

As previously demonstrated in level gait (walking and running), a polar logarithm graph is a valid tool to visualize the pattern of both amplitudes and phases.

The qualitative approach has shown that in gradient gaits:

- there are very few differences between males and females;
- as far as symmetrical coefficients have been regarded, we have found only little discrepancies as a function of gradient. Precisely, the most evident differences are at the highest slopes (both downhill and uphill);
- however, as far as asymmetrical coefficients have been regarded, we have found very anomalous patterns. This is probably related to the limited number of subjects who completed all the velocities in the protocol test;
- furthermore, whereas there are differences among gradients, these will be more referred to the pattern of phase instead of the amplitude.

In the following tables (males, Table a; and females, Table b), the main peculiarities of each coefficient as a function of slope (in males and females, walking and running) have been presented. All the observations refer to the level condition, in the same age group (25 to 35 years). Corresponding graphs have been presented in the paragraph above.

All these polar graphs are contained in the enclosed CD (First Study, Chapter 9, Polar log graphs: as a function of speed; as a function of gradient and with standard deviation).

In **gradient walking**:

A. Compared to the level gait, in the symmetrical coefficients (x2, y2 and z1) our results show that:

SYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x2 (forward)	very similar	very similar	very similar	very similar	very similar
y2 (vertical)	range from 1.50 to 0.83π	range from 1.33 to 0.83π	range from 1.33 to 1.00π	range from 1.50 to 1.00π	very similar
z1 (lateral)	very similar	very similar	very similar	very similar	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x2	very similar	very similar	very similar	very similar	very similar
y2	very similar	very similar	stops at 1.17π	very similar	very similar
z1	very similar	very similar	very similar	very similar	very similar

Table 9.9a. x2, y2 and z1 pattern in all walking gradients (males).

SYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x2 (forward)	data approaching 1.83π	slightly downward shift	quite similar	more confined range	slightly upward right shift
y2 (vertical)	range from 1.17 to 0.67π	range from 1.17 to 0.83π	range from 1.17 to 0.83π	range from 1.17 to 1.00π	quite similar
z1 (lateral)	very similar	very similar	very similar	very similar	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x2	very similar	very similar	very similar	stops at 1.50π	very similar
y2	very similar	very similar	wider range	very similar	very similar
z1	very similar	very similar	very similar	very similar	very similar

Table 9.9b. x2, y2 and z1 pattern in all walking gradients (females).

B. Compared to the level gait, in the other symmetrical coefficients (x4, y4 and z3) our results show that:

SYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x4 (forward)	quite similar	range from 1.83 to 1.33π	range from 1.83 to 1.50π	stops at 1.50π	very similar
y4 (vertical)	range from 1.00 to 0.33π	range from 1.00 to 0.33π	range from 1.00 to 0.33π	range from 1.17 to 0.17π	very similar
z3 (lateral)	quite similar	stops at 1.17π	very similar	stops at 1.17π	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x4	very similar	starts from 1.00π	starts from 1.00π	very similar	quite similar
y4	very similar	slightly downward shift	slightly downward shift	slightly downward shift	quite similar
z3	upward shift	quite similar	very similar	very similar	upward shift

Table 9.10a. x4, y4 and z3 pattern in all walking gradients (males).

SYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x4 (forward)	range from 1.67 to 1.00π	range from 1.67 to 1.17π	slightly downward shift	slightly downward shift	quite similar
y4 (vertical)	wider range	quite similar	quite similar	quite similar	anomalous pattern
z3 (lateral)	slightly downward shift	slightly downward shift	wider range	slightly downward shift	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x4	very similar	very similar	wider range	quite similar	slightly downward shift
y4	quite similar	anomalous pattern	anomalous pattern	anomalous pattern	slightly downward shift
z3	very similar	quite similar	quite similar	more confined range	quite similar

Table 9.10b. x4, y4 and z3 pattern in all walking gradients (females).

C. Compared to the level gait, in the asymmetrical coefficients (x1, y1 and z2) our results show that:

ASYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x1 (forward)	very similar	quite similar	data approaching 1.83π	quite similar	quite similar
y1 (vertical)	anomalous pattern	anomalous pattern	anomalous pattern	anomalous pattern	anomalous pattern
z2 (lateral)	very similar	anomalous pattern	shift to the opposite section	shift to the opposite section	anomalous pattern
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x1	quite similar	range from 1.67 to 1.17π	quite similar	anomalous pattern	anomalous pattern
y1	quite similar	shift to the opposite section	quite similar	anomalous pattern	quite similar
z2	anomalous pattern	quite similar	slightly right shift	anomalous pattern	wider range

Table 9.11a. x1, y1 and z2 pattern in all walking gradients (males).

ASYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x1 (forward)	wider range	quite similar	anomalous pattern	anomalous pattern	quite similar
y1 (vertical)	slightly upward shift	anomalous pattern	quite similar	very similar	very similar
z2 (lateral)	anomalous pattern	slightly downward shift	anomalous pattern	quite similar	quite similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x1	range from 0.83 to 1.17π	anomalous pattern	anomalous pattern	anomalous pattern	anomalous pattern
y1	upward shift	range from 0 to 0.33π	anomalous pattern	quite similar	anomalous pattern
z2	quite similar	quite similar	anomalous pattern	anomalous pattern	anomalous pattern

Table 9.11b. x1, y1 and z2 pattern in all walking gradients (females).

D. Compared to the level gait, in the other asymmetrical coefficients (x3, y3 and z4) our results show that:

ASYMMETRICAL COEFFICIENTS	-25% (MALES)	-20%	-15%	-10%	-5%
x3 (forward)	horizontal shift	anomalous pattern	anomalous pattern	shift to the opposite section	very similar
y3 (vertical)	shift to the upward section	anomalous pattern	anomalous pattern	quite similar	quite similar
z4 (lateral)	quite similar	shift to the opposite section	quite similar	very similar	quite similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x3	very similar	shift to the opposite section	range from 1.33 to 0.50π	anomalous pattern	anomalous pattern
y3	shift to the right section	quite similar	quite similar	quite similar	quite similar
z4	shift to the opposite section	quite similar	quite similar	quite similar	shift to the opposite section

Table 9.12a. x3, y3 and z4 pattern in all walking gradients (males).

ASYMMETRICAL COEFFICIENTS	-25% (FEMALES)	-20%	-15%	-10%	-5%
x3 (forward)	range from 1.17 to 1.67π	quite similar	quite similar	shift to the downward section	quite similar
y3 (vertical)	quite similar	very similar	very similar	quite similar	quite similar
z4 (lateral)	anomalous pattern	very similar	quite similar	very similar	quite similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x3	anomalous pattern	quite similar	quite similar	quite similar	quite similar
y3	anomalous pattern	quite similar	anomalous pattern	anomalous pattern	anomalous pattern
z4	quite similar	quite similar	quite similar	very similar	shift to the downward section

Table 9.12b. x3, y3 and z4 pattern in all walking gradients (females).

In **gradient running**:

A. Compared to the level gait, in the symmetrical coefficients (x2, y2 and z1) our results show that:

SYMMETRICAL COEFFICIENTS	-25% (MALES)	-20%	-15%	-10%	-5%
x2 (forward)	data approaching 0π	data approaching 0π	range from 1.83 to 1.00π	quite similar	very similar
y2 (vertical)	range from 0.83 to 0.50π	range from 0.67 to 0.50π	left shift	quite similar	very similar
z1 (lateral)	very similar	very similar	very similar	very similar	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x2	very similar	very similar	very similar	range from 0 to 1.83π	range from 1.83 to 1.50π
y2	quite similar (stops at 0π)	quite similar (stops at 0π)	range from 0.17 to 0π	range from 0.17 to 0π	range from 0.17 to 0π
z1	very similar	very similar	very similar	very similar	very similar

Table 9.13a. x2, y2 and z1 pattern in all running gradients (males).

SYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x2 (forward)	range from 0 to 1.83π	data approaching 0π	quite similar	data approaching 0.83π	very similar
y2 (vertical)	data approaching 0.50π	data approaching 0.50π	range from 0.33 to 0.50π	data approaching 0.33π	wider range
z1 (lateral)	very similar	very similar	very similar	very similar	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x2	quite similar	quite similar	data approaching 1.67π	range from 0 to 1.83π	data approaching 1.83π
y2	quite similar	data approaching 0.17π	very similar	more confined range	shift to the downward section
z1	very similar	very similar	very similar	very similar	very similar

Table 9.13b. x2, y2 and z1 pattern in all running gradients (females).

B. Compared to the level gait, in the other symmetrical coefficients (x4, y4 and z3) our results show that:

SYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x4 (forward)	range from 0.17 to 1.00π	range from 0.83 to 0.67π	range from 1.00 to 0.67π	slightly left shift	very similar
y4 (vertical)	right shift	range from 0.17 to 1.83π	range from 0.17 to 1.83π	range from 1.83 to 1.50π	quite similar
z3 (lateral)	range from 1.00 to 0.67π	range from 1.00 to 0.83π	quite similar	quite similar	very similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x4	stops at 0π	very similar	more confined range	very similar	very similar
y4	stops at 0.67π	shift to the upward section	stops at 0π	range from 1.17 to 0.33π	anomalous pattern
z3	very similar	stops at 0π	very similar	very similar	very similar

Table 9.14a. x4, y4 and z3 pattern in all running gradients (males).

SYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x4 (forward)	data approaching 0.50π	anomalous pattern	range from 0.33 to 0.83π	quite similar	very similar
y4 (vertical)	range from 0 to 1.67π	range from 0 to 1.67π	slightly right shift	range from 1.33 to 1.67π	quite similar
z3 (lateral)	data approaching 0.83π	range from 0.67 to 1.17π	wider range	wider range	quite similar
SYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x4	quite similar	range from 0.33 to 0π	quite similar	quite similar	quite similar
y4	range from 0.17 to 0π	wider range	quite similar	anomalous pattern	anomalous pattern
z3	anomalous pattern	anomalous pattern	wider range	anomalous pattern	range from 0 to 1.83π

Table 9.14b. x4, y4 and z3 pattern in all running gradients (females).

C. Compared to the level gait, in the asymmetrical coefficients (x1, y1 and z2) our results show that:

ASYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x1 (forward)	data approaching 0π	range from 1.67 to 1.83π	data approaching 1.50π	quite similar	quite similar
y1 (vertical)	anomalous pattern	range from 0.83 to 1.17π	quite similar	quite similar	very similar
z2 (lateral)	range from 1.00 to 1.50π	shift to the downward section	data approaching 0π	quite similar	very similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x1	very similar	very similar	quite similar	quite similar	quite similar
y1	quite similar	quite similar	quite similar	anomalous pattern	anomalous pattern
z2	quite similar	quite similar	range from 0.83 to 0.67π	quite similar	quite similar

Table 9.15a. x1, y1 and z2 pattern in all running gradients (males).

ASYMMETRICAL COEFFICIENTS	**-25% (FEMALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x1 (forward)	quite similar	quite similar	shift to the upward section	shift to the right upward section	very similar
y1 (vertical)	quite similar	quite similar	quite similar	very similar	very similar
z2 (lateral)	anomalous pattern	left shift	shift to the downward section	slightly left shift	slightly left shift
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x1	anomalous pattern	quite similar	quite similar	anomalous pattern	range from 1.50 to 1.83π
y1	anomalous pattern	shift to the opposite section	anomalous pattern	wider range	anomalous pattern
z2	quite similar	quite similar	quite similar	quite similar	anomalous pattern

Table 9.15b. x1, y1 and z2 pattern in all running gradients (females).

D. Compared to the level gait, in the other asymmetrical coefficients (x3, y3 and z4) our results show that:

ASYMMETRICAL COEFFICIENTS	**-25% (MALES)**	**-20%**	**-15%**	**-10%**	**-5%**
x3 (forward)	quite similar	anomalous pattern	shift to the upward section	wider range	anomalous pattern
y3 (vertical)	anomalous pattern	anomalous pattern	quite similar	quite similar	quite similar
z4 (lateral)	quite similar	right shift	shift to the downward section	quite similar	very similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x3	quite similar	range from 1.00 to 0.50π	anomalous pattern	quite similar	right shift
y3	quite similar	quite similar	very similar	more confined range	anomalous pattern
z4	quite similar	quite similar	quite very similar	very similar	quite similar

Table 9.16a. x3, y3 and z4 pattern in all running gradients (males).

ASYMMETRICAL COEFFICIENTS	-25% (FEMALES)	-20%	-15%	-10%	-5%
x3 (forward)	shift to the downward section	range from 1.67 to 1.00π	quite similar	quite similar	shift to the left downward section
y3 (vertical)	anomalous pattern	quite similar	quite similar	anomalous pattern	anomalous pattern
z4 (lateral)	quite similar	very similar	very similar	very similar	very similar
ASYMMETRICAL	**5%**	**10%**	**15%**	**20%**	**25%**
x3	quite similar	quite similar	quite similar	range from 0.67 to 1.00π	quite similar
y3	anomalous pattern	quite similar	quite similar	quite similar	anomalous pattern
z4	very similar	very similar	very similar	anomalous pattern	quite similar

Table 9.16b. x3, y3 and z4 pattern in all running gradients (females).

5. CONCLUSION

The graphical approach starting from Fourier analysis has gone to polar logarithm graphs.

In detail, they have been in characterizing a successful solution which simultaneously represents the amplitude (A) and the phase (φ) pattern. Indeed, they permit to better visualize them in a definite geometrical form (i.e. the circumference).

Moreover, the calculation of the standard deviations has completed our outlook to this method.

In our discussion, only a restricted number of such examples has been proposed for subjects (males and females) aged 25 to 35. Specifically, this age group seems to present the most regular and clear pattern compared to the other ages.

This constitutes the most important result underlying that young adults move (walk and run) in the more appropriate, constant, normal and expected way.

Furthermore, it has been demonstrated that gender and gradient don't play a crucial role while both age and speed do.

In all harmonic coefficients, relevant graphical differences have been found among gaits.

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Chapter 10

BIOMECHANICAL VARIABLES IN WALKING AND RUNNING

1. INTRODUCTION

1.1. Main biomechanical variables

In this chapter, we will focus on some simple biomechanical variables (Taylor, 1994) (stride frequency and stride length - par. 2, 3 and 4 - and duty factor - par. 5 -) and complex biomechanical variables (mechanical external work - par. 6 -, energy recovery percentage - par. 7 -, mechanical internal work - par. 8 - and mechanical total work - par. 9 -) which are important to extract and characterize the individual gait signature. Therefore, knowing these biomechanical variables becomes fundamental both to fully describe the mechanics of walking and running (see also chapter 1, par. 4) and to extract and characterize the individual gait signature (see also chapter 7).

Each biomechanical variable has been elaborated by means of a custom-written LabVIEW software (Minetti et al., 1993; see also chapter 6, par. 2). As a result, these kinematic and biomechanical variables were measured discretely cycle by cycle, at the chosen sampling rate (100 Hz; see also chapter 4, par. 3.4.4), in order to finally obtain average values. These values are then related to each testing condition: walking and running as gender, age, speed and gradient change.

Single values of simple and complex biomechanical variable are contained in the enclosed CD (First Study, Chapter 10, Biomechanical variables in level and gradient gaits - both in males and females -).

1.2. Statistical analysis

Statistical analysis was performed by using each subject biomechanical variable value.

Results will be presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The chosen independent variables were age group (y), progression speed (m/s) and gradient (%). The dependent variables were the stride frequency (SF), the stride length (SL), the duty factor (DF), the mechanical external work (W_{ext}), the energy recovery percentage (R), the mechanical internal work (W_{int}) and the mechanical total work (W_{tot}).

Effects of gender and age on each dependent variable were assessed by using a one-way ANOVA for unrelated measures. In addition, a post-*hoc* Bonferroni test was used to detect the strength of the associations between each dependent variable and gender/age.

Moreover, effects of speed and gradient were assessed by using a one-way ANOVA for related measures. In addition, a post-*hoc* paired *t*-test (with Bonferroni correction) was used to detect differences between each dependent variable and speed/gradient.

SPSS software (version 12.0 for Windows) was used for statistical analysis (Zakeri et al., 2006; Houdijk et al., 2009).

Specific results of this statistical analysis are presented and described in the Appendix 10.1.

Specifically, single values of biomechanical variables are contained in the enclosed CD (First Study, Chapter 10, Statistical analysis).

1.3. Graph legend

In each graph, the points represent mean values obtained by grouping the same age subjects in the different testing condition. The lines represent the simple graphic amalgamation of all the data; the vertical bars represent positive and negative standard deviations of the higher and lower speed curves (mean ± S.D.), respectively.

The graph legend is the same as already illustrated and described in chapter 8 (par. 3).

2. STRIDE FREQUENCY

2.1. Introduction

A person constrained to move at a given speed *s* on a treadmill chooses a particular stride frequency and stride length (see par. 3 below). Testing over a range of speeds generates a speed-frequency relationship (Unnithan et al., 1990; Bertram et al., 2001). This relationship is commonly posited as a basic feature of human gait. Therefore, stride frequency (SF) is the number of strides performed within a limited time period or one second (Du Chatinier et al., 1970; Cavagna et al., 1986; Laurent et al., 1986; Nilsson et al., 1987; Zatsiorsky et al., 1994; Farley et al., 1996; Kang et al., 2002; Danion et al., 2003; Chau et al., 2004; Pachi, 2005; Holt et al., 2006; Hoyt et al., 2006; Lippert, 2006; Segers et al., 2006; Grimshaw et al., 2007; Richards, 1999; De Smet et al., 2009). Walking humans prefer to use the stride rate that results in the lowest rate of metabolic energy expenditure (Minetti et al., 1993; Umberger et al., 2007).

Since it is expressed as a frequency (Hz), this parameter is the reciprocal of this time period or the inverted value of the stride cycle time (Stokes et al., 1998; Korhonen et al., 2009). So that, SF=1/T, where *T* is the time between two successive foot contacts (Segers et al., 2006).

Referring to our data, stride frequency has been calculated as:

$$SF = \frac{1}{\frac{(\text{endframetime} - \text{startframetime})}{\text{stridenumber}}} \quad [\text{Eq. } 10.1]$$

where *end frame time* is the last good (correct marker position recording) one; *start frame time* is the first good one; and *stride number* corresponds to the strides performed within the limited period defined by start and end frames. The knowledge of end frame, start frame and stride number comes from kinematic data analysis (see also chapter 6, par. 2.1).

2.2. Stride frequency as a function of age

2.2.1. Stride frequency in level walking

Precisely, our results show that:

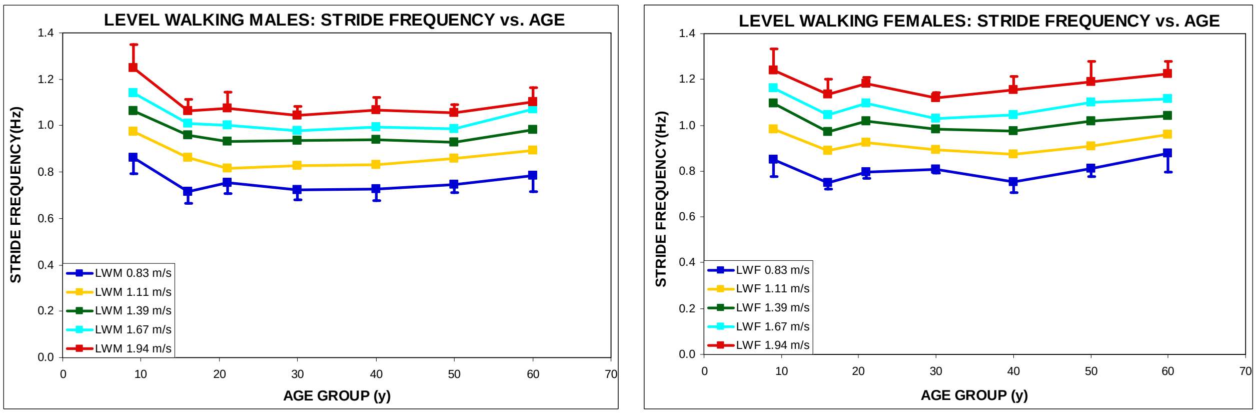


Figure 10.1. Stride frequency as a function of age in level walking, males (on the left) and females (on the right).

- stride frequency is higher in young male (1.058 ± 0.075 Hz, at all the investigated speeds) and female (1.067 ± 0.073 Hz) children aged 6 to 13. This is due to the fact of their shorter anthropometric dimensions (Grieve et al., 1966; Hoffmann, 1971; Rosenrot et al., 1980; Sutherland et al., 1980; Gatesy et al., 1991; Yamasaki et al., 1991; Jeng et al., 1997; Gurney, 2002; Kang et al., 2002; Kimura et al., 2009; Korhonen et al., 2009);
- there is no significant relationship comparing other age groups (0.919 ± 0.053 Hz independently of age and speed, in males; and 0.990 ± 0.040 Hz, in females; Cunningham et al., 1982; Hamilton, 1993);
- as expected, this pattern is similar at each speed;

- however, in females (right graph), only the qualitative analysis shows that stride frequency little increases with age (Hageman et al., 1986; Yamasaki et al., 1991; Barak et al., 2006; Mian et al., 2006; Kimura et al., 2007; Ortega et al., 2007; Cavagna et al., 2008b). This is probably due to the fact of the articular (Grieve et al., 1966; Hageman et al., 1986) and muscular (Narici et al., 2003; Reeves et al., 2003a; 2003b) constraints in elderly subjects;
- all our results concur with literature data (Grieve et al., 1966; Hageman et al., 1986; Hamilton, 1993; Schepens et al., 1998; 2001; Kang et al., 2002; Danion et al., 2003; Mian et al., 2006; Cavagna et al., 2008b; Kimura et al., 2009; Korhonen et al., 2009).

Specific results of the statistical analysis (with relevance) are shown in par. 2.1, Table 10.1 (see Appendix 10.1). Furthermore, the qualitative analysis shows that, on average, stride frequency is slightly greater in females (1.001 ± 0.045 Hz, independently of age and speed) than in males (0.965 ± 0.060 Hz, according to Kang et al., 2002).

2.2.2. Stride frequency in level running

Precisely, our results show that:

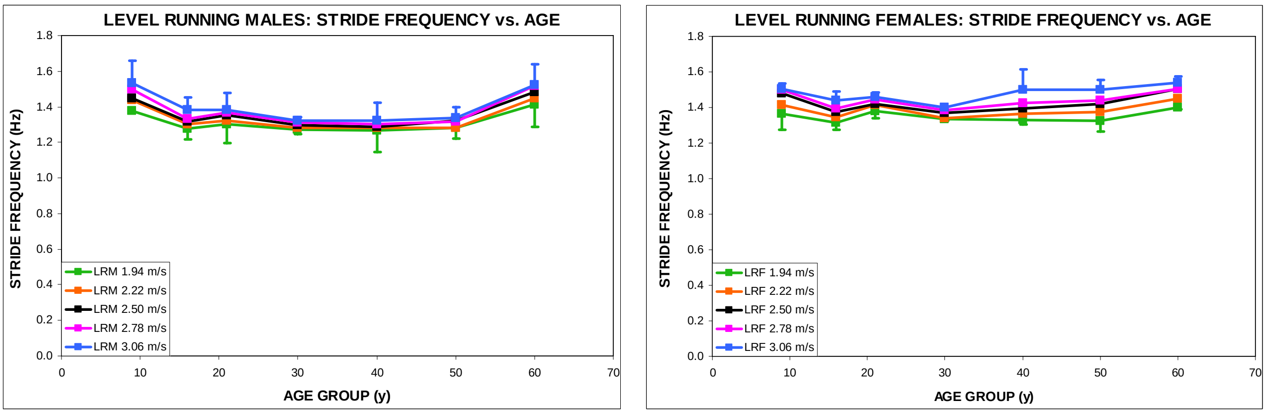


Figure 10.2. Stride frequency as a function of age in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), there is no significant variability among all age groups. This pattern is similar at each speed;
- all our results concur with literature data (Hoffmann, 1971; Hageman et al., 1986; Unnithan et al., 1990; Hamilton, 1993; Schepens et al., 1998; 2001; Danion et al., 2003; Korhonen et al., 2003; Kimura et al., 2007; Cavagna et al., 2008b: Figure 10.3; Fukuchi et al., 2008; Kimura et al., 2009; Korhonen et al., 2009).

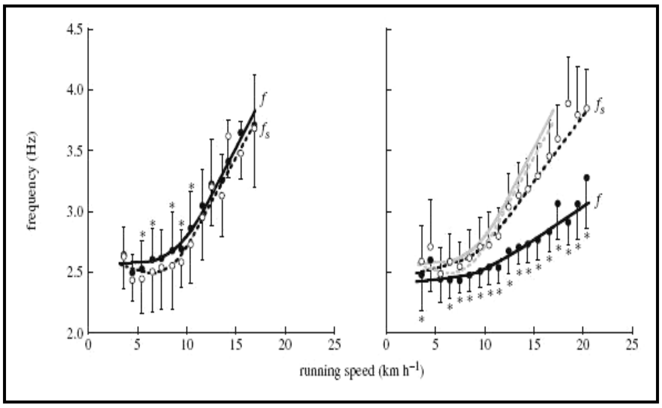


Figure 10.3. SF as a function of age and speed, in Cavagna et al. (2008b).
The step frequency (continuous line) is greater in the old subjects (73.6 ± 5.5 years, on the left) than in the young subjects (20.8 ± 1.6 years, on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 2.2, Table 10.2 (see Appendix 10.1). Furthermore, the qualitative analysis shows that, stride frequency is little greater in females (1.416 ± 0.050 Hz, independently of age and speed) than in males (1.357 ± 0.082 Hz). This pattern occurs especially in subjects aged 46 to 65: 1.445 ± 0.048 Hz, in females, at all the investigated speeds; and 1.392 ± 0.094 Hz, in males.

2.3. Stride frequency as a function of speed

2.3.1. Stride frequency in level walking and running

Stride frequency in walking is highly different to stride frequency in running (p<0.001).

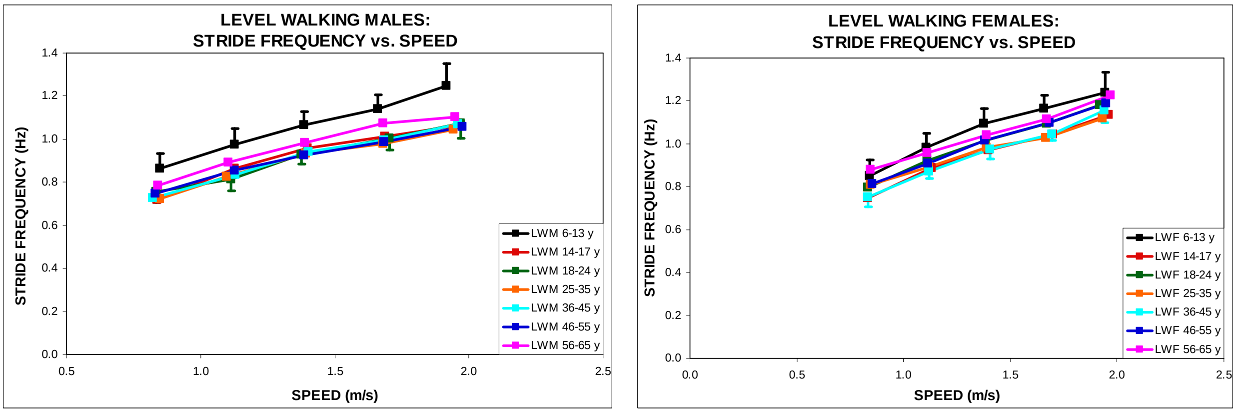


Figure 10.4. Stride frequency as a function of speed in level walking,
males (on the left) and females (on the right).

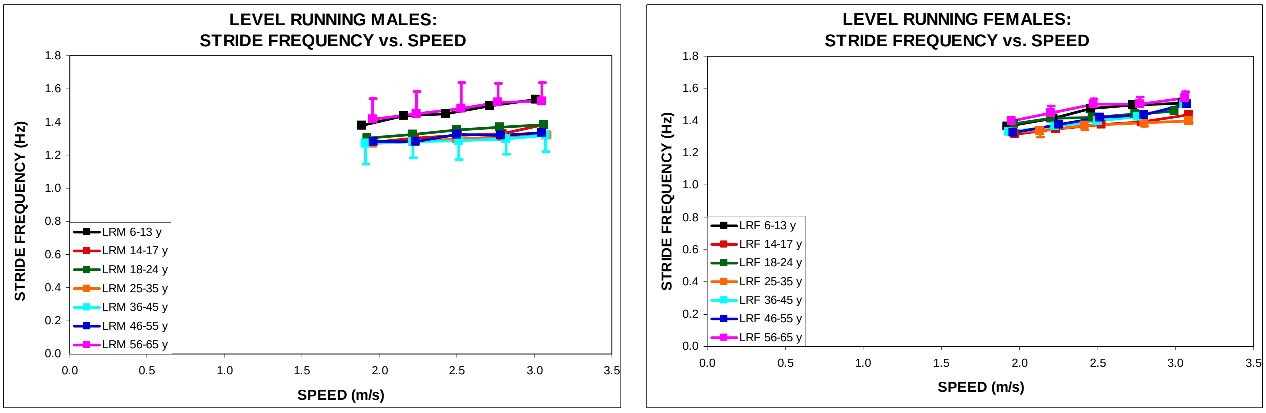


Figure 10.5. Stride frequency as a function of speed in level running, males (on the left) and females (on the right).

In level walking (Figure 10.4), our results show that there is an upward linear function with speed at all ages regardless of gender ($p<0.001$ both in males and females). However, in level running (Figure 10.5), our results show that step frequency is about constant ($p<0.05$).

All these results concur with the documented data (Grieve et al., 1966; Hoffmann, 1971; Cavagna et al., 1976; 1977; Rosenrot et al., 1980; Cavagna et al., 1988; Steudel, 1990b; Lejeune et al., 1998; Minetti, 1998; Schepens et al., 1998; Minetti et al., 2001b; Schepens et al., 2001; Danion et al., 2003; Mercer et al., 2003; Saibene et al., 2003; Herbin et al., 2004; Bereket, 2005; Bertram, 2005; Ortega et al., 2005; Biewener, 2006; Hoyt et al., 2006; Witte et al., 2006; Morin et al., 2007; Rowlands et al., 2007; Thomas et al., 2007; Osaki et al., 2008; Kavanagh, 2009).

Specific results of the statistical analysis (with relevance) are shown in par. 2.3, Table 10.3 (see Appendix 10.1).

2.3.2. Abrupt change in stride frequency in gait transition

For a variable to be considered a gait transition, it has been reasoned that: a) there would be an increase in the value of the variable as the walking speed increased, but an abrupt change in the value of the variable as the gait changed from walk to run; b) the transition would always occur at the same critical value of the variable; and c) the variable would have the potential to influence the proprioceptive feedback gained during the movement (Hreljac, 1995a).

Consequently, in stride frequency, there is an abrupt change in gait transition (from walking to running; Hreljac, 1993a; 1993b; 1995a; 1995b; Minetti et al., 2001b; Raynor et al., 2002; Day et al., 2006; Hoyt et al., 2006; Segers et al., 2006), as shown in Figure 10.6.

This pattern occurs both in males (left graph) and females (right graph).

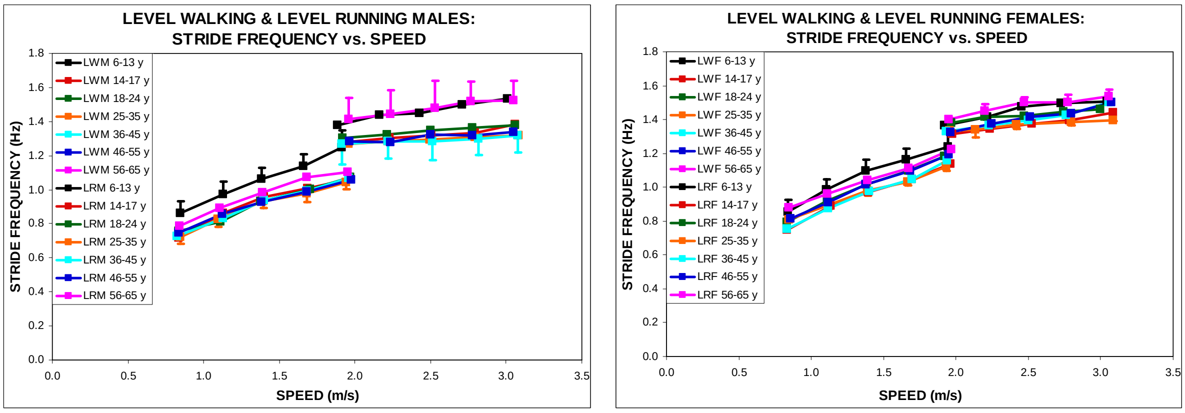


Figure 10.6. The abrupt change in gait transition in stride frequency,
males (on the left) and females (on the right).

As shown in the graphs below, all our results concur with literature data (Minetti, 1998: Figure 10.7, left graph; Saibene et al., 2003: Figure 10.7, right graph).

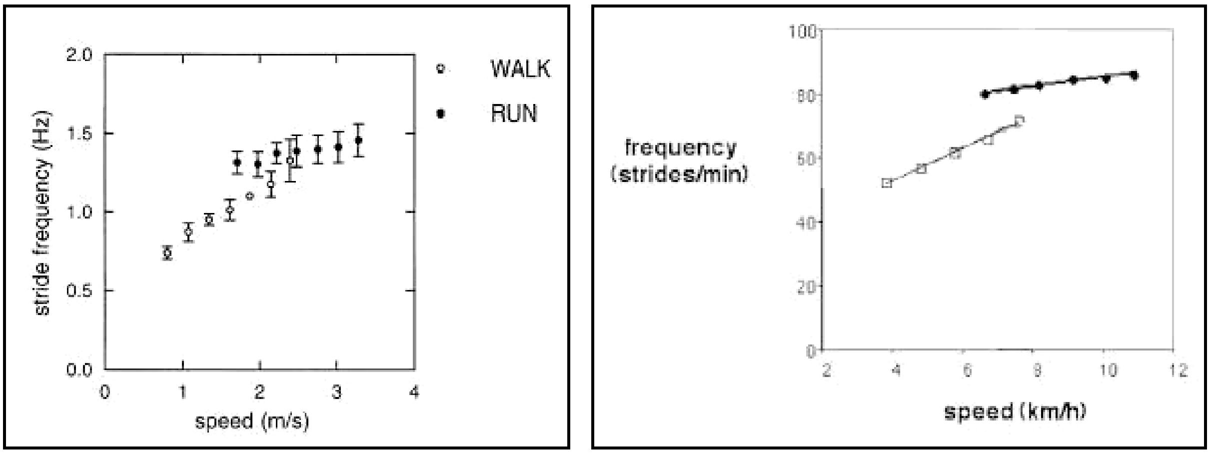


Figure 10.7. SF as a function of speed, in Minetti (1998) (on the left)
and in Saibene et al. (2003) (on the right).

2.4. Stride frequency as a function of gradient

For males and females aged 25 to 35, we analysed stride frequency in relation to gradient, too (Figure 10.8 and 10.9).

2.4.1. Stride frequency in gradient walking

Precisely, both in males (left graph) and females (right graph), our results show that:

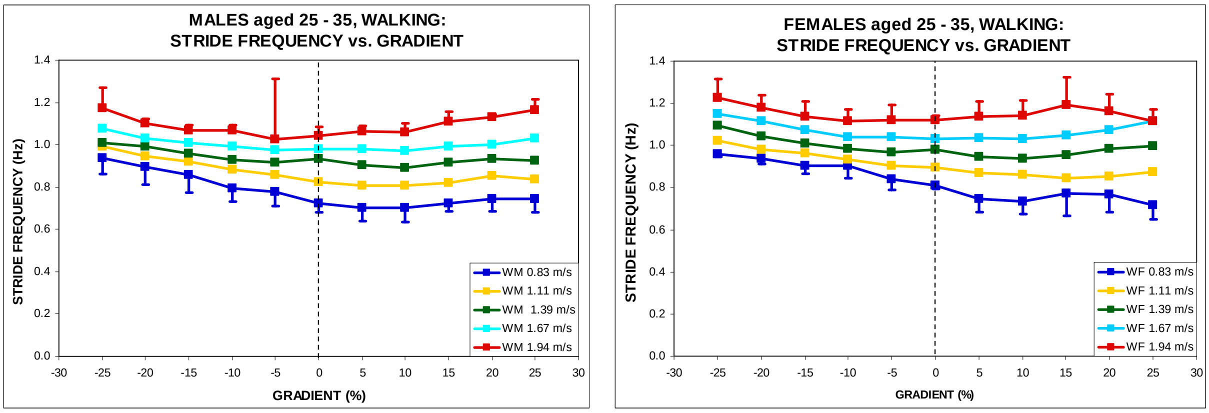


Figure 10.8. Stride frequency as a function of gradient in walking,
males (on the left) and females (on the right).

- at the different speeds, stride frequency slightly decreases with gradient from -25% to the level condition (from 1.037 ± 0.070 to 0.901 ± 0.044 Hz, at all the investigated speeds, in males; and from 1.090 ± 0.061 to 0.967 ± 0.022 Hz, in females). As expected, this pattern is more evident at the lower speeds (0.83 m/s, $p<0.001$; and 1.11 m/s, $p<0.05$, respectively);
- furthermore, in general, at each speed, there is no significant change in stride frequency from the level condition to the maximum upgrade (from 0.901 ± 0.044 to 0.940 ± 0.051 Hz, at all the investigated speeds, in males; and from 0.967 ± 0.022 to 0.964 ± 0.097 Hz, in females, according to Kang et al., 2002);
- in males, only at the higher speed (1.94 m/s), stride frequency little increases from 10 to 15% (from 1.058 ± 0.046 to 1.110 ± 0.045 Hz, $p<0.001$).

Specific results of the statistical analysis (with relevance) are shown in par. 2.4, Table 10.4 (see Appendix 10.1).

2.4.2. Stride frequency in gradient running

Precisely, both in males (left graph) and females (right graph), our results show that:

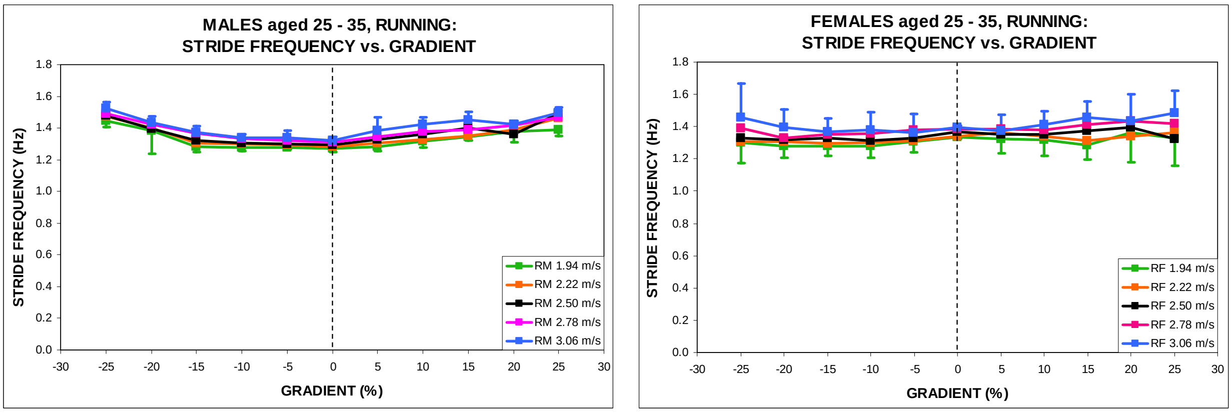


Figure 10.9. Stride frequency as a function of gradient in running,

males (on the left) and females (on the right).

- at different speeds, there is no significant change in stride frequency regarding gradient (Minetti et al., 1994: Figure 10.10; Swanson et al., 2000);
- this pattern is quite similar at each speed.

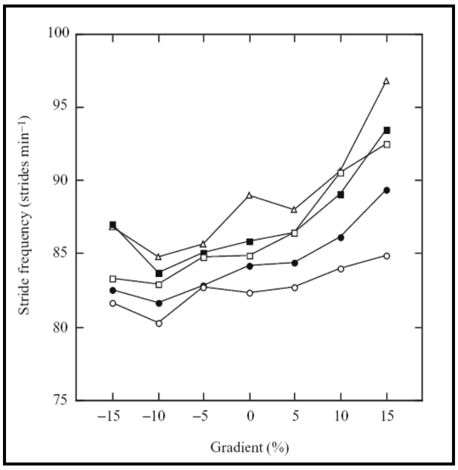


Figure 10.10. SF at different speeds as a function of gradient, in Minetti et al. (1994).

Specific results of the statistical analysis (with relevance) are shown in par. 2.5, Table 10.5 (see Appendix 10.1).

3. STRIDE LENGTH

3.1. Introduction

Stride length estimation is an important issue in areas such as gait analysis, sport training or pedestrian localization (Lippert, 2006; Gonzalez et al., 2007).

Stride length (SL) is the distance between corresponding points on successive footprints of the same foot (Du Chatinier et al., 1970; Afelt et al., 1983; Laurent et al., 1986; Nilsson et al., 1987; Winter et al., 1990; Alexander, 1992; Ostrosky et al., 1994; Zatsiorsky et al., 1994; Judge et al., 1996a; 1996b; Griffin et al., 1999; Richards, 1999; Sparrow, 2000; Rowlands et al., 2001; Kang et al., 2002; Mercer et al., 2002; Danion et al., 2003; Holt et al., 2006; Segers et al., 2006; Witte et al., 2006; Grimshaw et al., 2007; Reisman et al., 2007; Wu et al., 2007; De Smet et al., 2009; ESMAC Hand Notes, 2009; Franz et al., 2009; Korhonen et al., 2009; Montero-Odasso et al., 2009; Racic et al., 2009; van de Walle et al., 2009).

It has been calculated as:

$$SL = \frac{s}{SF} \quad [\text{Eq. } 10.2]$$

where *s* is the average progression speed (m/s); and *SF* is the average stride frequency (Hz). Therefore, the stride length is expressed in m (Unnithan et al., 1990).

3.2. Results of our experiments

Because of the strong relationship among stride frequency and stride length (see par. 3.1 above), for SL we applied the same statistical analysis previously used for SF, as well. Consequently, in both stride frequency and stride length, significances and results are clearly the same.

In the Appendix 10.1, the only stride frequency statistical results are presented in order to avoid repetitions.

3.2.1. Stride length as a function of age

As shown in Figure 10.11, our results show that:

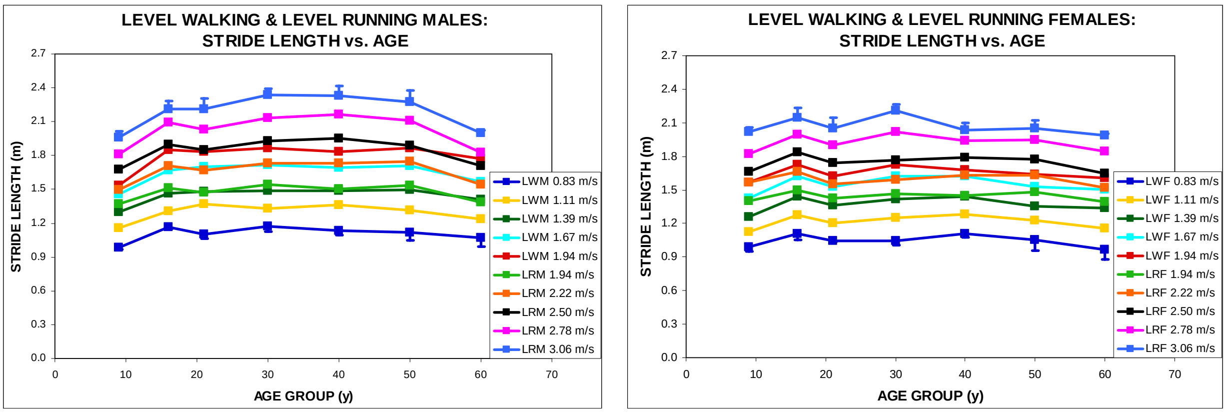


Figure 10.11. Stride length as a function of age in level walking and running, males (on the left) and females (on the right).

- as expected, in walking, stride length is lower in young male (1.286 ± 0.223 m, at all the investigated speeds) and female (1.272 ± 0.231 m) children aged 6 to 13. This is due to their shorter anthropometric dimensions (Grieve et al., 1966; Himann et al., 1988; Elble et al., 1991; Gatesy et al., 1991; Yamasaki et al., 1991; Hamilton, 1993; Jeng et al., 1997; Sekiya et al., 1997; Hoyt et al., 2000; Varraine et al., 2000; Gurney, 2002; Kang et al., 2002; Bastien et al., 2003; Lippert, 2006; Kimura et al., 2009; Korhonen et al., 2009);
- there is no significant relationship comparing other age groups (1.486 ± 0.281 m, independently of age and speed, in males; and 1.383 ± 0.348 m, in females; Himann et al., 1988; Yamasaki et al., 1991; Hamilton, 1993; Gurney, 2002; Kimura et al., 2007; Ortega et al., 2007; Hernandez et al., 2009). However, our result seems to be partially in contrast with Winter et al. (1990), Ostrosky et al. (1994), Judge et al. (1996a), Korhonen et al. (2003), Hausdorff (2004), Fukuchi et al. (2008) and Korhonen et al. (2009). Particularly, it has been

verified that older people had a 10% shorter step length during usual gait, when corrected for leg length (Judge et al., 1996a). Furthermore, in another study it has been demonstrated that stride length showed clear reductions with increasing age (Korhonen et al., 2003). This discrepancy is probably due to the different ranges of age investigated: from 6 to 65 years in our study *versus* from 35 to 88 years in other researches;

- this pattern is similar at each speed;
- however, in running, both in males and females, there is no significant variability among all age groups (Unnithan et al., 1990). This pattern is similar at each speed;
- importantly, on average, stride length is slightly higher in males (1.437 ± 0.420 m, independently of age and speed) than in females (1.367 ± 0.245 m; Kang et al., 2002; Barret et al., 2008);
- all our results concur with literature data (Grieve et al., 1966; Himann et al., 1988; Hamilton, 1993; Elble et al., 1991; Alexander, 1992; Sekiya et al., 1997; Hoyt et al., 2000; Kang et al., 2002; Danion et al., 2003; Barak et al., 2006; Kimura et al., 2007; 2009).

3.2.2. Stride length as a function of speed

Our manner of movement changes (i.e. stride length) with speed (Alexander, 1992). Particularly, both in walking and in running, stride length increases with speed (Boje, 1944; Grieve et al., 1966; Zarruch et al., 1974; Cavanagh et al., 1982; Varraine et al., 2000; Donelan et al., 2002b; Alexander, 2004; Paroczai et al., 2006; Neptune et al., 2008; Franz et al., 2009).

Stride length in walking is highly different to stride length in running (p<0.001).

As shown in Figure 10.12, our results show that:

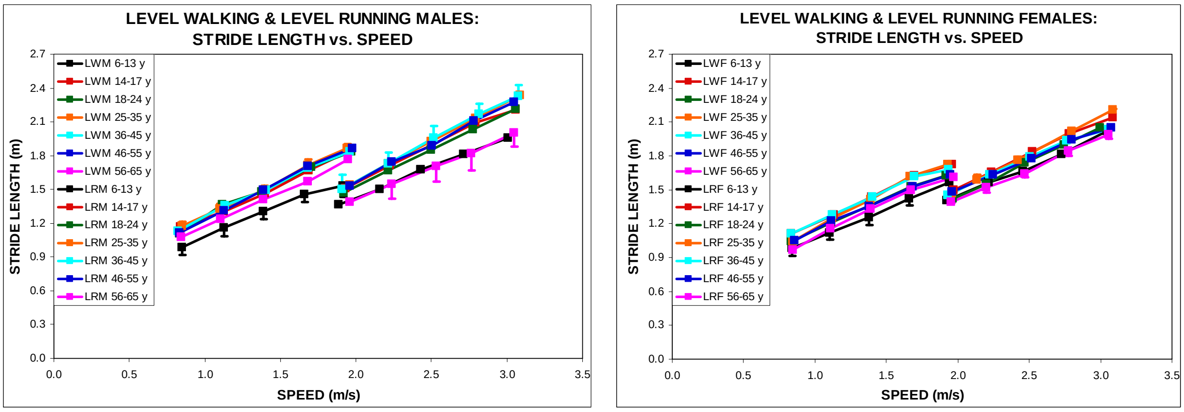


Figure 10.12. Stride length as a function of speed in level walking and running, males (on the left) and females (on the right).

- there is an upward linear function with speed (Grieve et al., 1966; Derrick et al., 1998; Hoyt et al., 2000; Mercer et al., 2003; Herbin et al., 2004; Bereket, 2005; Bertram, 2005; Thomas et al., 2007; Barret et al., 2008; Franz et al., 2009; Kavanagh, 2009) at all ages regardless of gender ($p<0.001$ both in males (left graph) and females (right graph));
- in stride length, there is an abrupt change in gait transition (from walking to running; Hreljac, 1993a; 1993b; 1995a; 1995b; Raynor et al., 2002; Day et al., 2006; Segers et al., 2006). This pattern occurs both in males and females.

3.2.3. Stride length as a function of gradient

As shown in Figure 10.13, our results show that:

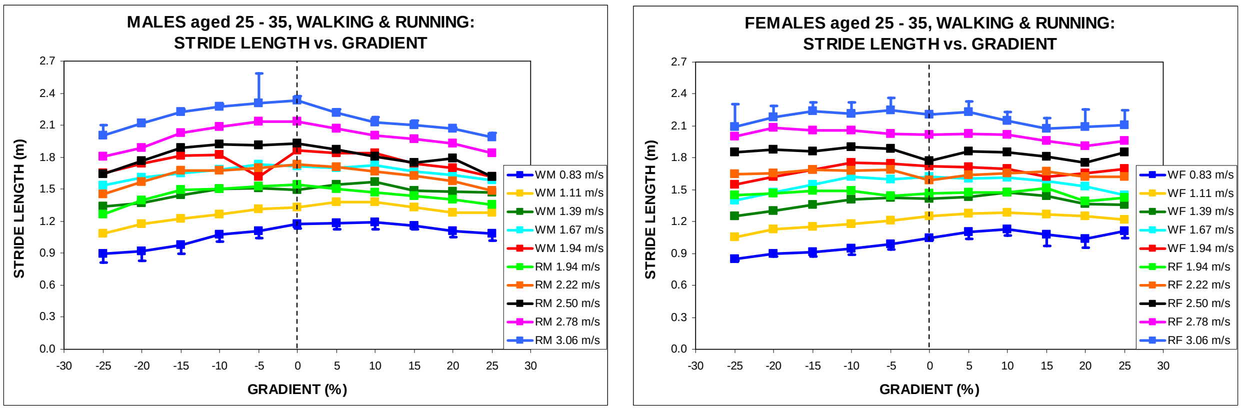


Figure 10.13. Stride length as a function of gradient in walking and running, males (on the left) and females (on the right).

In gradient walking, both in males (left graph) and females (right graph), our results show that:

- at the different speeds, stride length slightly increases with gradient from -25% to the level condition (from 1.299 ± 0.312 to 1.514 ± 0.279 m, at all the investigated speeds, in males; and from 1.217 ± 0.277 to 1.411 ± 0.273 m, in females);
- as expected, this pattern is more evident at the lower speeds (0.83 m/s, $p<0.001$; and 1.11 m/s, $p<0.05$);
- furthermore, in general, at each speed, there is no significant change in stride length from the level condition to the maximum upgrade (Kang et al., 2002).

In gradient running, both in males (left graph) and females (right graph), our results show that:

- at different speeds, there is no significant change in stride length regarding gradient (Rowlands et al., 2001). This pattern is similar at each speed.

4. THE RELATIONSHIP AMONG STRIDE FREQUENCY AND STRIDE LENGTH

4.1. Walking gait

The combined effects of stride frequency (shaded squares) and stride length (blank squares) on walking speed are illustrated in Figure 10.14 both in males (left graph) and females (right graph).

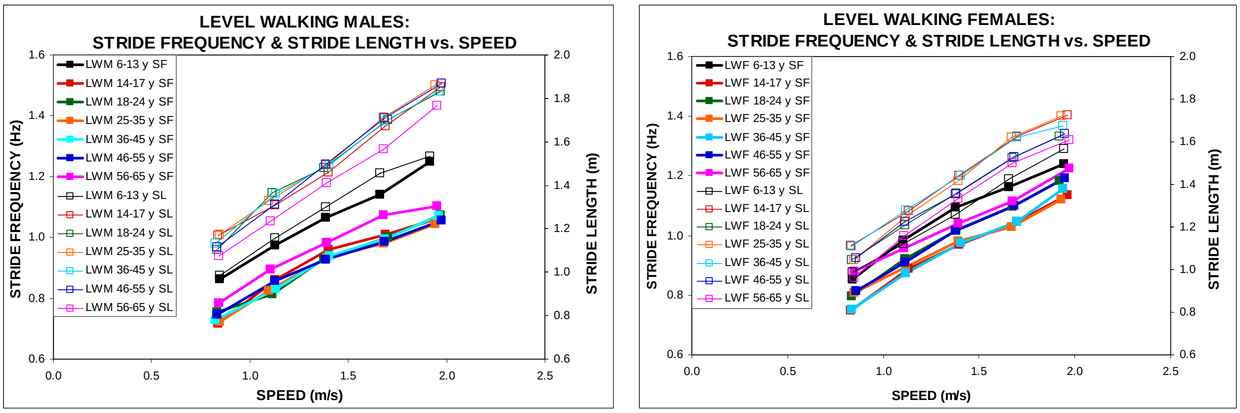


Figure 10.14. The relationships among SF, SL and walking speed,
males (on the left) and females (on the right).

Particularly, both stride frequency and stride length increases linearly with walking speed as already illustrated and described in par. 2 and 3 (Bonnard et al., 1993; Kuo, 2001; Danion et al., 2003; Usherwood et al., 2008; De Smet et al., 2009).

Moreover, it is important to note that within certain limits there is a similarity in average values of stride frequency when males are compared to females, and *vice versa*.

However, average values of stride length are little higher in males than in females (see par. 3.2 above). Finally, in males both stride frequency and stride length vary in a bigger range of value.

4.2. Running gait

Running speed depends on two variables, stride frequency and stride length (Laurent et al., 1986; Enoka, 2002). As widely demonstrated in literature (Unnithan et al., 1990; Bonnard et al., 1993; McGinnis, 2005; De Smet et al., 2009), if stride length remains constant, then as stride frequency increases running speed increases; moreover, if stride frequency remains constant, speed increases as stride length increases.

The combined effects of stride frequency and stride length on running speed are illustrated in Figure 10.15 both for males (left graph) and females (right graph).

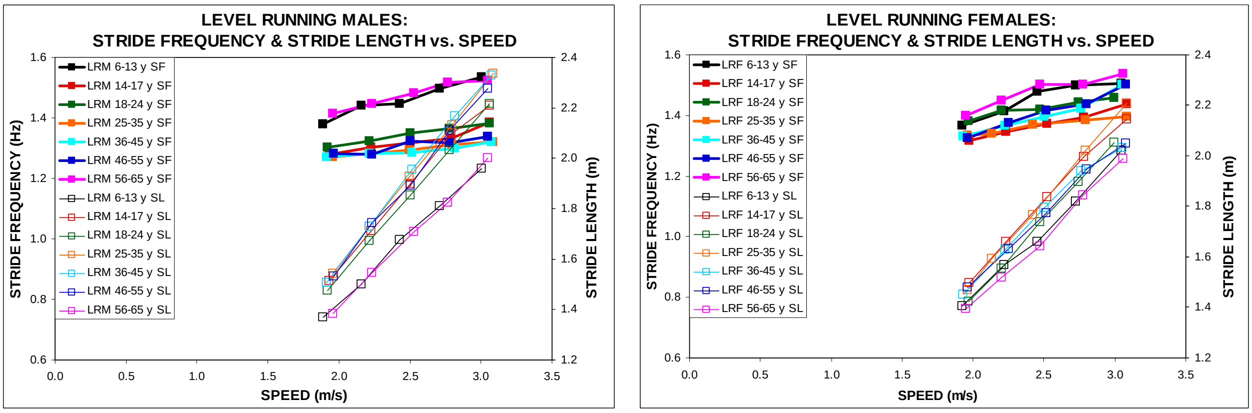


Figure 10.15. The relationships among SF, SL and running speed, males (on the left) and females (on the right).

Particularly, both stride frequency and stride length increases linearly with running speed as already illustrated and described in par. 2 and 3. However, it is important to note that the increase in stride length is higher than that one in stride frequency. In fact, the contribution of changes of these variables are different at low and high velocities: this is apparent by the differences in the slope of each curve at different velocities (Unnithan et al., 1990; Bonnard et al., 1993; Enoka, 2002).

Finally, notice that stride frequency increases continually with speed (the slope is not that steep) both at lower and higher velocities speeds; however, stride length increases with speed only at lower speeds (as accurately described by De Smet et al., 2009, in the analysis of the choice of the desired speed by a runner). Moreover, within certain limits there is a similarity in average values of stride frequency when males are compared to females, and *vice versa*. However, average values of stride length are little higher in males than in females (the slope is steeper).

5. DUTY FACTOR

5.1. Introduction

Duty factor (DF) is the fraction of the duration of the stride period when each foot is on the ground or the time of contact relative to entire stride cycle duration (%contact; Alexander, 1989; Ettema, 1996; Minetti et al., 1997; Griffin et al., 1999; Alexander, 2004; Biewener, 2006; Biknevicius et al., 2006; Bullimore et al., 2006; Hoyt et al., 2006; Segers et al., 2006; Witte et al., 2006; Segers et al., 2007a; 2007b; Korhonen et al., 2009; Ruckstuhl et al., 2009).

Many studies have demonstrated that duty factor is a key factor affecting both the energetics and mechanics of running (Morin et al., 2007; Ruckstuhl et al., 2009). It depends on the combined effects (Saibene et al., 2003) of: a) the stiffness of the leg (Farley et al., 1996; Arampatizs et al., 1999; Kerdok et al., 2002); b) the track (Ferris et al., 1998); c) the body mass of the subject; and d) the body size (Biewener, 1983).

In walking, duty factor is greater than 0.5 (‘pendulum mechanics’; see also chapter 1, par. 4.3); in running, however, is less than 0.5 so there are times when both feet are off the ground (‘spring mechanics’; see also chapter 1, par. 4.4) (Alexander, 1992; Donelan et al., 1997; Alexander, 2004; Biewener, 2006; Biknevicius et al., 2006; Hoyt et al., 2006; Segers et al., 2007a).

In our study, it is available directly from the custom-written LabVIEW software (Minetti et al., 1993).

5.2. Duty factor as a function of age

Duty factor is always higher in walking than in running (Minetti, 1998). In effect:

- in level walking, DF ranges from 48.900 ± 2.920 to 62.960 ± 1.316%, in males; and from 48.100 ± 1.980 to 64.280 ± 1.931%, in females;
- in level running, DF ranges from 27.140 ± 2.169 to 41.475 ± 4.263%, in males; and from 25.500 ± 0.283 to 43.875 ± 3.459%, in females.

5.2.1. Duty factor in level walking

Precisely, our results show that:

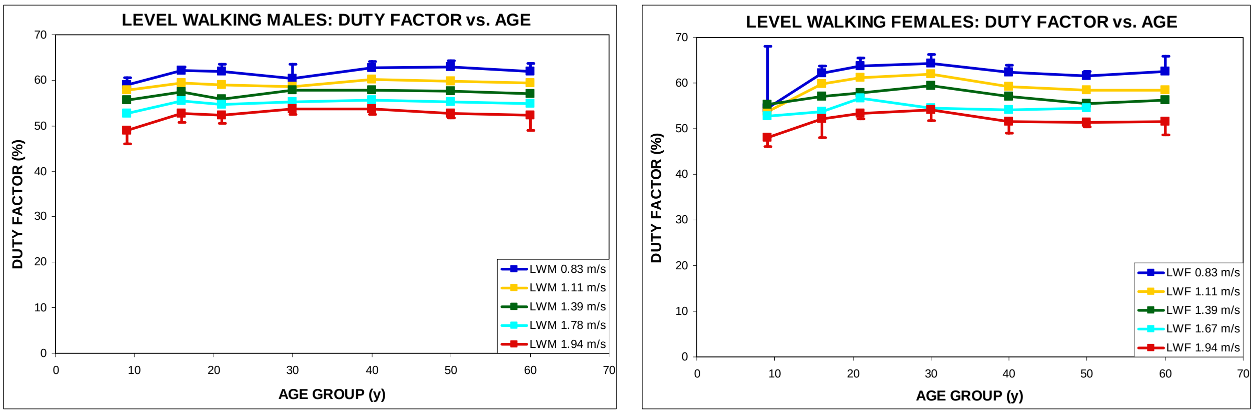


Figure 10.16. Duty factor as a function of age in level walking, males (on the left) and females (on the right).

- as expected, duty factor decreases as walking speed increases (see par. 5.3.1 below);
- in males (left graph), there is no significance between duty factor and age. This pattern occurs at each speed;
- however, in young female children aged 6 to 13 (right graph), at the highest speed (1.94 m/s), duty factor is slightly lower (54.100 ± 2.395%, p<0.001). This is probably due to their lowest anthropometric body mass dimensions (Minetti et al., 1992);
- at the other speeds, in females, there is no significance between duty factor and age (58.039 ± 2.625%, independently of age and speed);

- all our results concur with literature data (Cavagna et al., 1983; 1986; Winter et al., 1990; Gatesy et al., 1991; Hamilton, 1993; Sparrow, 2000; Korhonen et al., 2003; Cavagna et al., 2008a; Korhonen et al., 2009).

Specific results of the statistical analysis (with relevance) are shown in par. 3.1, Table 10.6 (see Appendix 10.1).

5.2.2. Duty factor in level running

Precisely, our results show that:

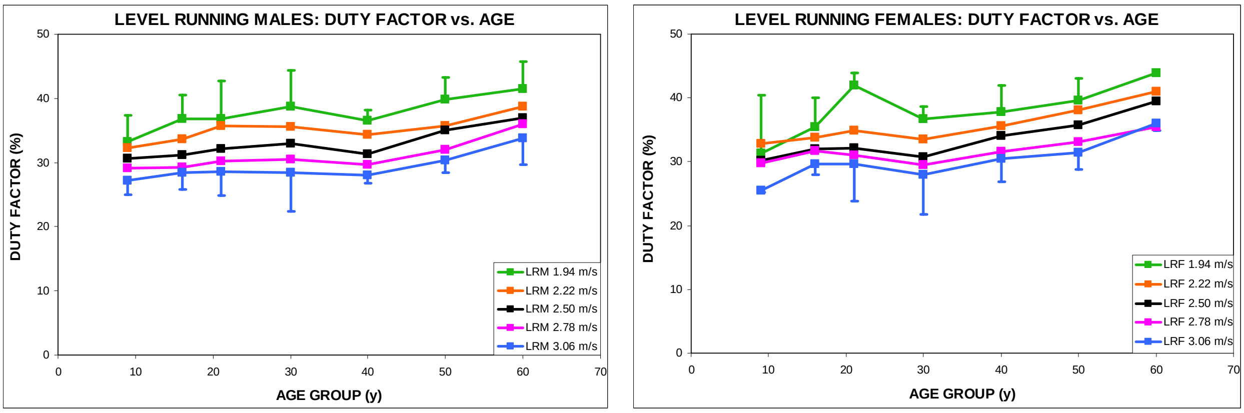


Figure 10.17. Duty factor as a function of age in level running, males (on the left) and females (on the right).

- as expected, duty factor decreases as running speed increases (see par. 5.3.2 below);
- both in males (left graph) and females (right graph), there is no significant variability among all age groups. This pattern is similar at each speed;
- all our results concur with literature data (Winter et al., 1990; Gatesy et al., 1991; Hamilton, 1993; Sparrow, 2000; Korhonen et al., 2003; Cavagna et al., 2008a; 2008b; Korhonen et al., 2009).

Specific results of the statistical analysis (with relevance) are shown in par. 3.2, Table 10.7 (see Appendix 10.1).

5.3. Duty factor as a function of speed

5.3.1. Duty factor in level walking

Both in level walking (Figure 10.18) and running (Figure 10.19), there is a downward linear function with speed at all ages regardless of gender ($p<0.001$ both in males (left graph) and females (right graph)) (Alexander, 1989; Minetti, 1998; Schepens et al., 1998; Alexander, 2004; Biewener, 2006; Witte et al., 2006; Morin et al., 2007).

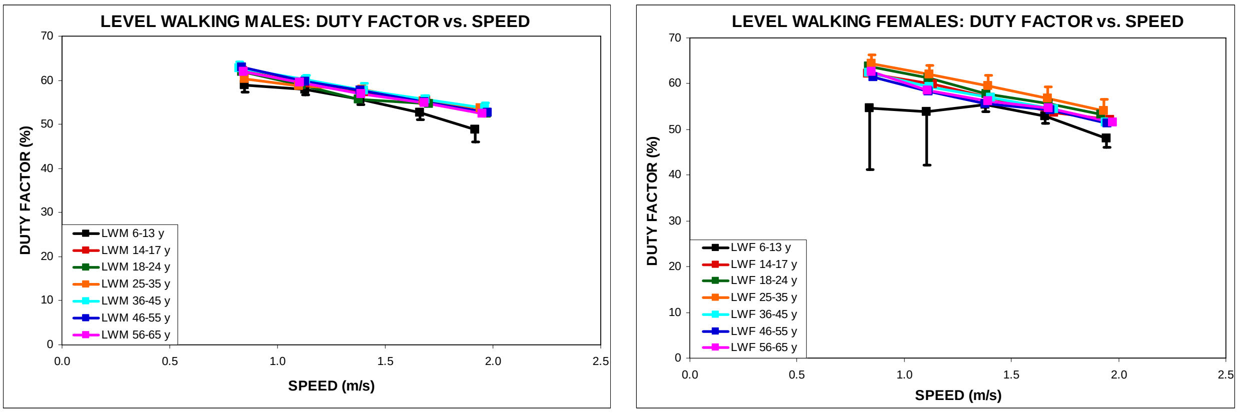


Figure 10.18. Duty factor as a function of speed in level walking, males (on the left) and females (on the right).

Furthermore, as previously described:

- in level walking, duty factor is lower in male (54.804 ± 1.764%, at all the investigated speeds) and female (52.928 ± 6.013%) children aged 6 to 13;
- moreover, in young females aged 6 to 13, duty factor presents a non-regular pattern: it little increases with speed up to 1.39 m/s (from 54.560 ± 13.387 to 55.360 ± 1.582%, $p<0.05$), and then it decreases with speed (from 55.360 ± 1.582 to 48.100 ± 1.980%, $p<0.001$).

Specific results of the statistical analysis (with relevance) are shown in par. 3.3, Table 10.8 (see Appendix 10.1).

5.3.2. Duty factor in level running

As previously described, in level running:

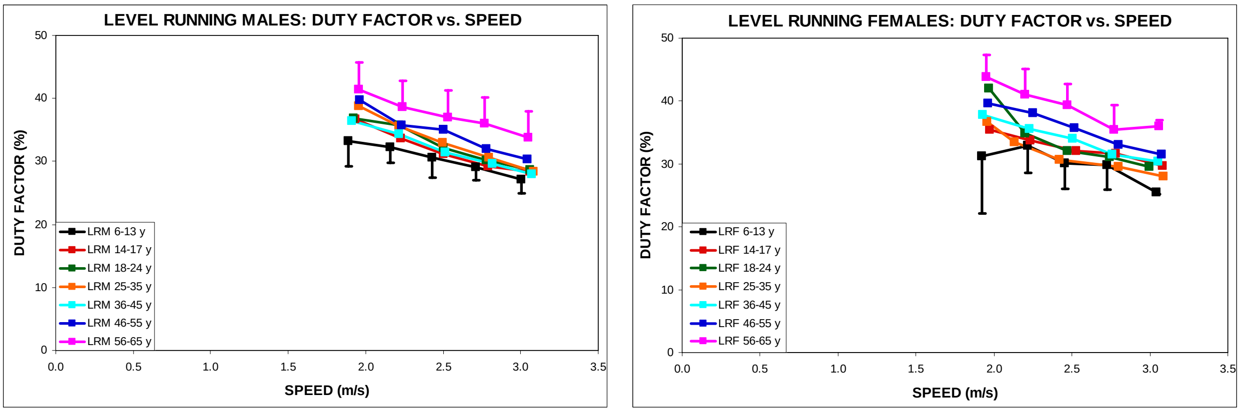


Figure 10.19. Duty factor as a function of speed in level running, males (on the left) and females (on the right).

- duty factor is lower in young male (30.480 ± 2.803%, at all the investigated speeds) and female (29.916 ± 4.343%) children aged 6 to 13;

- however, it is greater in male (35.988 ± 3.480%, at all the investigated speeds) and female (37.377 ± 3.027%) adults aged 46 to 65. This pattern could be probably due to a reduced stiffness in tendons in elderly adults as demonstrated in Hubbard et al. (1984), Reeves et al. (2003a), Karamanidis et al. (2005), Narici et al. (2005) and Magnusson et al. (2008);
- moreover, in young females aged 6 to 13, duty factor presents a non-regular pattern: it increases with speed up to 1.11 m/s (from 31.240 ± 9.135 to 32.860 ± 4.255%, $p<0.05$), and then it decreases with speed (from 32.860 ± 4.255 to 25.550 ± 0.283, $p<0.001$).

5.3.3. Abrupt change in duty factor in gait transition

Furthermore, in duty factor, there is an abrupt change (Figure 10.20) in gait transition (from walking to running; Alexander, 1989; Hreljac, 1993a; 1993b; 1995a; 1995b; Sparrow, 2000; Raynor et al., 2002; Day et al., 2006; Hoyt et al., 2006; Segers et al., 2006). This pattern occurs both in males (left graph) and females (right graph).

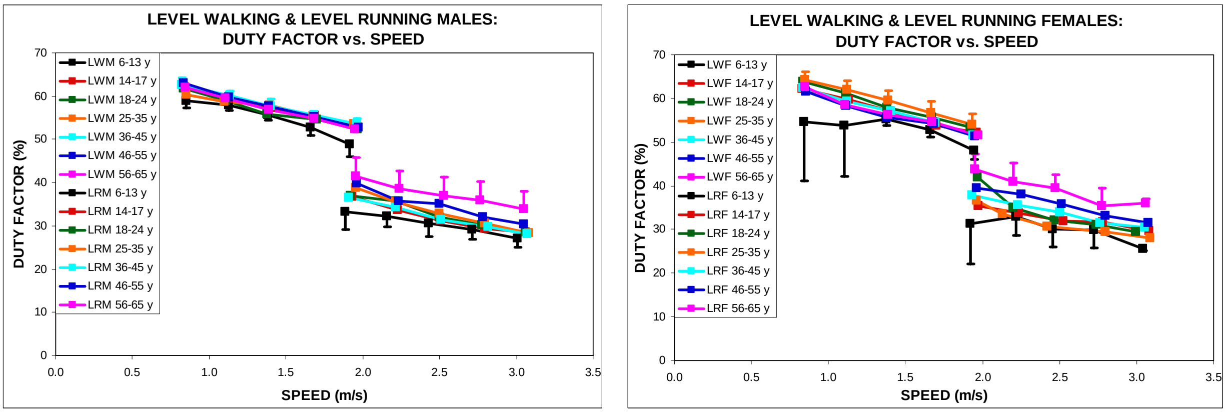
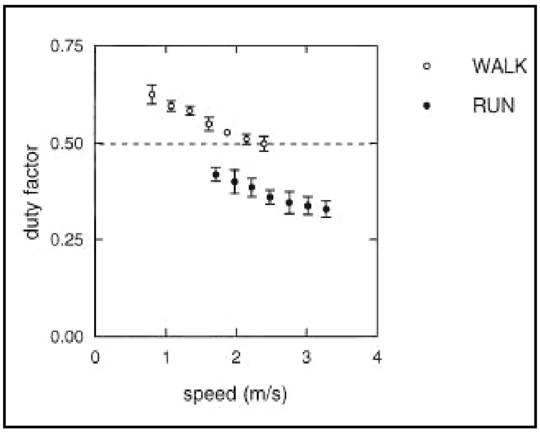


Figure 10.20. The abrupt change in gait transition in duty factor,
males (on the left) and females (on the right).



All our results concur with literature data (Minetti, 1998: Figure 10.21, DF as a function of speed; Alexander, 2004; Cavagna et al., 2008b).

5.4. Duty factor as a function of gradient

For males and females aged 25 to 35, we analysed duty factor in relation to gradient, too (Figure 10.22 and 10.23).

5.4.1. Duty factor in gradient walking

As expected, duty factor is always higher at the lower speeds (0.83 and 1.11 m/s).

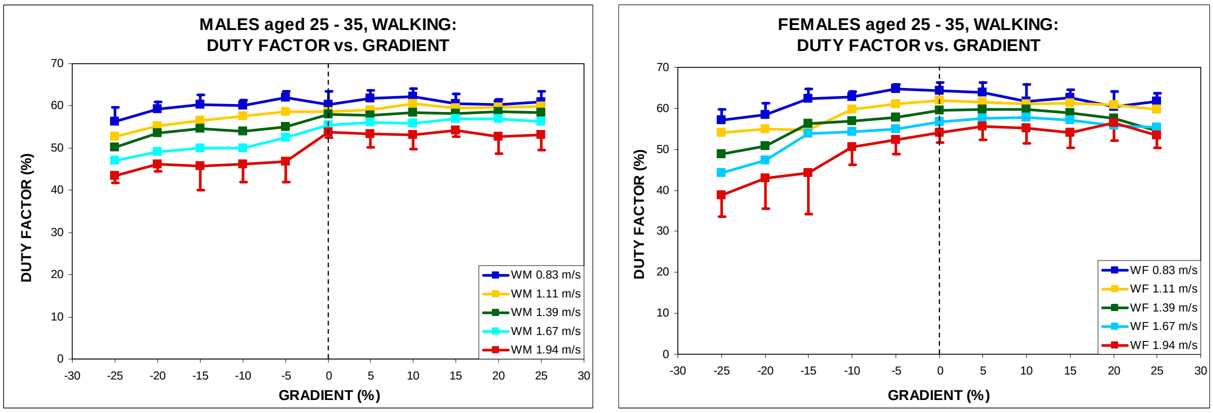


Figure 10.22. Duty factor as a function of gradient in walking, males (on the left) and females (on the right).

Precisely, in males (left graph), our results show that:

- duty factor does not significant change at the lower speeds (0.83 and 1.11 m/s);
- however, at the other speeds, there is a little upward linear function with gradient, from -25 to -20% (from 36.283 ± 9.254 to 47.075 ± 5.911%, at all the investigated speeds, $p<0.01$). Above this gradient, it does not significantly change;
- the only exception is at the higher speed of 1.94 m/s in which duty factor slightly increases from -25 to -15% (from 27.075 ± 12.450 to 45.775 ± 5.613%, $p<0.01$); it then little increases from -5% to the level condition (from 46.879 ± 14.963 to 53.675 ± 1.261%, $p<0.01$); finally, it does not change at the other gradients.

In females (right graph):

- in general, duty factor little significantly changes with gradient ($p<0.05$);
- this pattern is quite similar at each speed.

Specific results of the statistical analysis (with relevance) are shown in par. 3.4, Table 10.9 (see Appendix 10.1).

5.4.2. Duty factor in gradient running

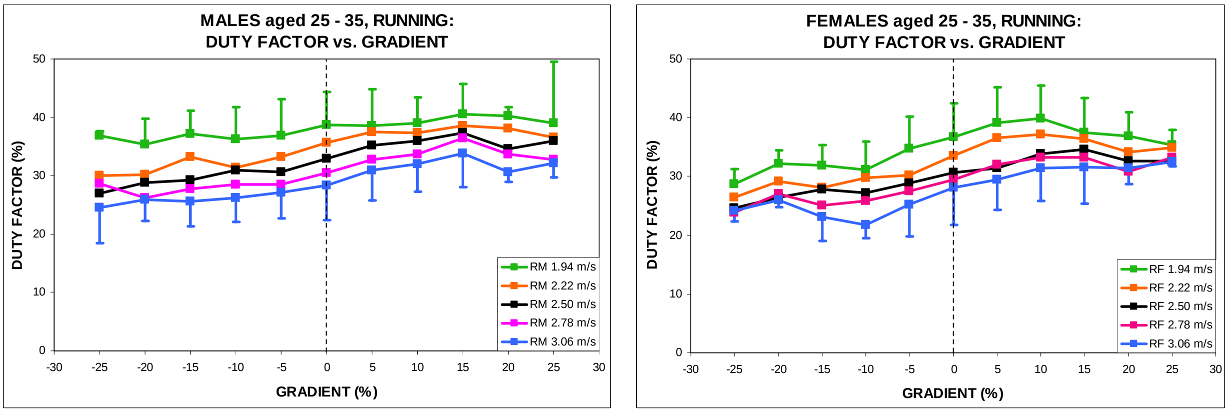


Figure 10.23. Duty factor as a function of gradient in running, males (on the left) and females (on the right).

Precisely, in males (left graph), our results show that:

- in general, duty factor does not significantly change with gradient (Minetti et al., 1994: Figure 10.24);
- this pattern is almost similar at each speed;
- the only exceptions are at the speeds of 2.22 and 3.06 m/s. Precisely, it increases from -10 to 10% (from 31.440 ± 4.968 to 37.380 ± 3.989% at 2.22 m/s and 26.280 ± 4.220 to 32.020 ± 4.747% at 3.06 m/s, $p<0.05$).

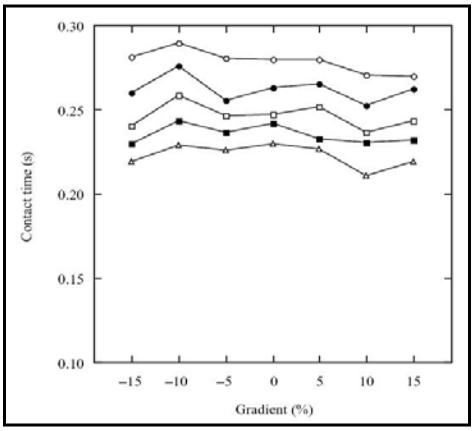


Figure 10.24. Contact time in a stride at different speeds as a function of gradient, in Minetti et al. (1994).

Specific results of the statistical analysis (with relevance) are shown in par. 3.5, Table 10.10 (see Appendix 10.1). Statistical analysis was not applied to female data, because of some discarded tests (see also chapters 5, par. 1.1.2 and 6, par. 2.2).

In females (right graph), the qualitative analysis shows that there is no significance between duty factor and gradient. This pattern occurs at each speed.

6. MECHANICAL EXTERNAL WORK

6.1. Mechanical external work definition

The study of external work has resulted in the identification of the two generally accepted fundamental mechanisms of terrestrial locomotion, in particular, the pendulum-like model of walking and the bouncing model of running, trotting and hopping (see also chapter 1, par. 4).

Particularly, mechanical external work (W_{ext}) accounts for the changes in potential (PE) and kinetic (KE) energies of the BCOM with respect to the environment (Cavagna et al., 1976; 1977; Aleshinsky, 1986b; Steudel, 1990b; Flynn et al., 1993; Sun et al., 1993; Willems et al., 1995; Duff-Raffaele et al., 1996; Lejeune et al., 1998; Minetti, 1998; Schepens et al., 1998; Cavagna et al., 2000; Cerretelli, 2001; Minetti et al., 2001a; 2001b; Schepens et al., 2001; Terrier et al., 2001; Donelan et al., 2002a; Kautz et al., 2002; Bastien et al., 2003; Hallemans et al., 2004; Minetti, 2004;

Ortega et al., 2005; Mian et al., 2006; Minetti et al., 2006; Devita et al., 2007; van de Hecke et al., 2007; Umberger et al., 2007; Mahaudens et al., 2008; Sasaki et al., 2008; Winiarski, 2008; Genin et al., 2009; Houdijk et al., 2009; Mahaudens et al., 2009; Malatesta et al., 2009; Peyrot et al., 2009; van de Walle et al., 2009).

Precisely, mechanical external work can be divided into:

1. W_{ext}^{+} corresponding to an increase in total mechanical energy: this positive mechanical work raises and accelerates the BCOM (Minetti et al., 1993; Sparrow, 2000; Donelan et al., 2002a; Devita et al., 2007; Umberger et al., 2007; Schepens et al., 2009);
2. W_{ext}^{-} corresponding to a decrease in total mechanical energy: this negative mechanical work lowers and decelerates the BCOM (Minetti et al., 1993; Sparrow, 2000; Donelan et al., 2002a; Devita et al., 2007; Umberger et al., 2007; Schepens et al., 2009).

It can be obtained both using dynamometric platforms (*direct dynamics*; Cavagna et al., 1963; Belli et al., 1993; Ozkaya et al., 1998; Schepens et al., 1998; 2001; Donelan et al., 2002a; Mahaudens et al., 2009; Schepens et al., 2009) and cinematographic data (*inverse dynamics*; Ardigò, 1992; Minetti et al., 1993; 1994; Malatesta et al., 2009; Peyrot et al., 2009). Recently, it has been measured by satellite positioning system, as well (Terrier et al., 2001).

It is also well known that, in level walking at a constant speed, the amount of positive mechanical work (W_{ext}^{+}), mostly due to ascending and accelerations of the BCOM, must be counterbalanced by an equal quantity of negative work (W_{ext}^{-}), related to descending and deceleration and braking effect (Williams et al., 1983; Sparrow, 2000; Donelan et al., 2002b; Umberger et al., 2007). It has also been shown that:

- W_{ext} reaches a highest value at the most economical speed of walking (Cavagna et al., 1963; Duff-Raffaele et al., 1996);
- at each gradient, there is a unique W_{ext}^{+}/W_{ext}^{-} ratio (= 1 in level walking), regardless of speed, with a tendency for W_{ext}^{+} and W_{ext}^{-} to disappear above +15% and below -15%, respectively (Ardigò, 1992; Minetti et al., 1993);
- at any given speed, W_{ext} is the smallest the higher the frequency used (Cavagna et al., 1986);
- furthermore, the average mechanical power (e.g. the positive work done at each step divided by the step period) is minimized at a step frequency close to the freely chosen step frequency (Duff-Raffaele et al., 1996; Cavagna et al., 1997; Umberger et al., 2007);
- the tendency of W_{ext} to increase at high frequencies is due to the persistent minimal vertical excursion of the BCOM (*the locomotory dead space*; Minetti et al., 1995);
- in old age, $W_{ext,}$ to maintain the motion of the BCOM, is reduced due to the lower resistance to gravity (Cavagna et al., 2008b).

For other information concerning this important complex biomechanical variable, see also chapters 1, 11 and 12.

6.2. Forward and vertical work in walking and running

Both in walking and running, we know that W_{ext} is less the sum of:

1. the absolute values of the work to lift the BCOM, W_v (Cavagna et al., 1976; Lejeune et al., 1998; Griffin et al., 1999; Schepens et al., 2001; Ivanenko et al., 2004; Ortega et al., 2007);
2. the absolute values of the work to accelerate the BCOM forward, W_f because of a transfer between gravitational potential and kinetic energy (Cavagna et al., 1976; Lejeune et al., 1998; Griffin et al., 1999; Schepens et al., 2001; Ivanenko et al., 2004).

W_v decreases with frequency similarly in children and adults, regardless of gender, age and body dimensions (Schepens et al., 1998).

All our results concur with literature data (Figure 10.25).

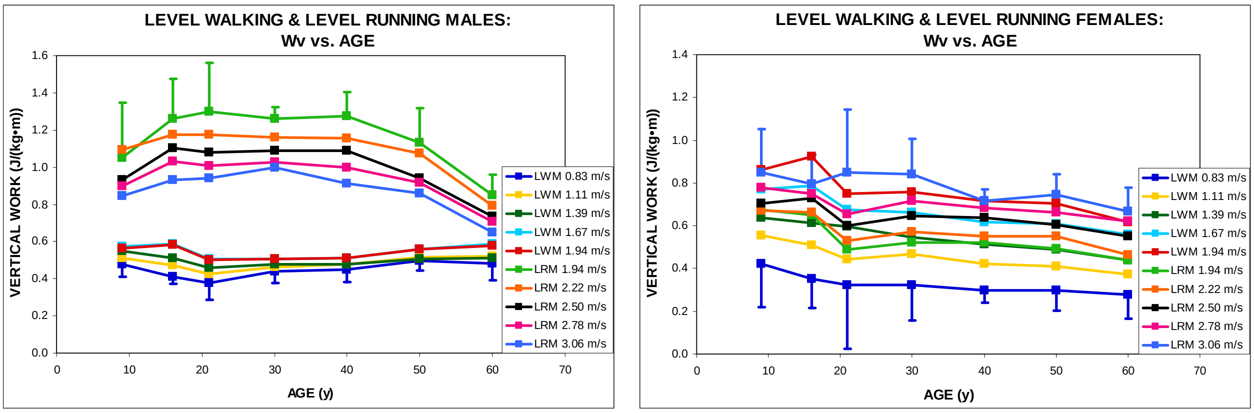


Figure 10.25. Vertical work as a function of age in level walking and running, males (on the left) and females (on the right).

On the other hand, W_f remains practically constant or increases with frequency in children, whereas it decreases slightly with frequency in adults: this difference becomes greater with increased speed (Schepens et al., 2001). All our results concur with literature data (Figure 10.29).

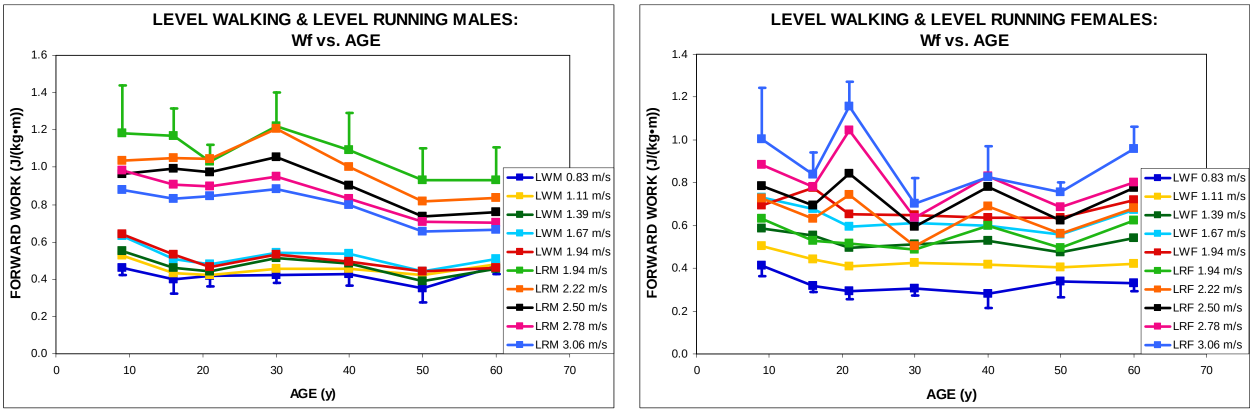


Figure 10.26. Forward work as a function of age in level walking and running, males (on the left) and females (on the right).

Moreover, it has been showed that whereas the W_f increases similarly with speed in old and young subjects (both in walking and running), the resistance to gravity W_v is appreciably lower the older the subject (Cavagna et al., 2008b).

Particularly (Figure 10.27 and 10.28), W_v is similar to W_f at the intermediate speed (≈ 1.39 m/s). At the lower speeds (0.83 and 1.11 m/s), W_v is greater than W_f; however, at the higher speeds (1.67 and 1.94 m/s), W_v is lower than W_f (Saibene et al., 2003).

Finally, there is a downward linear relation between W_v (and W_f) and speed in level running (both in males and females; Willems et al., 1995).

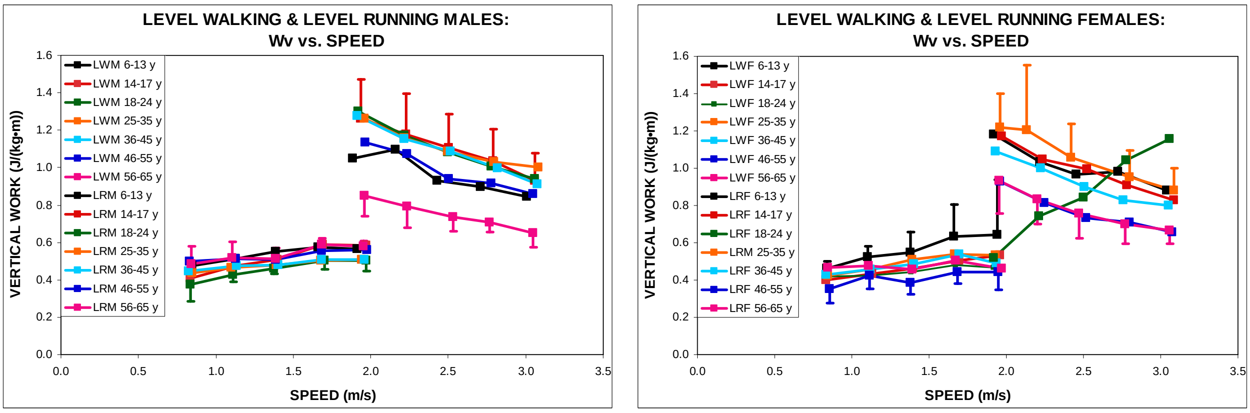


Figure 10.27. Vertical work as a function of speed in level walking and running, males (on the left) and females (on the right).

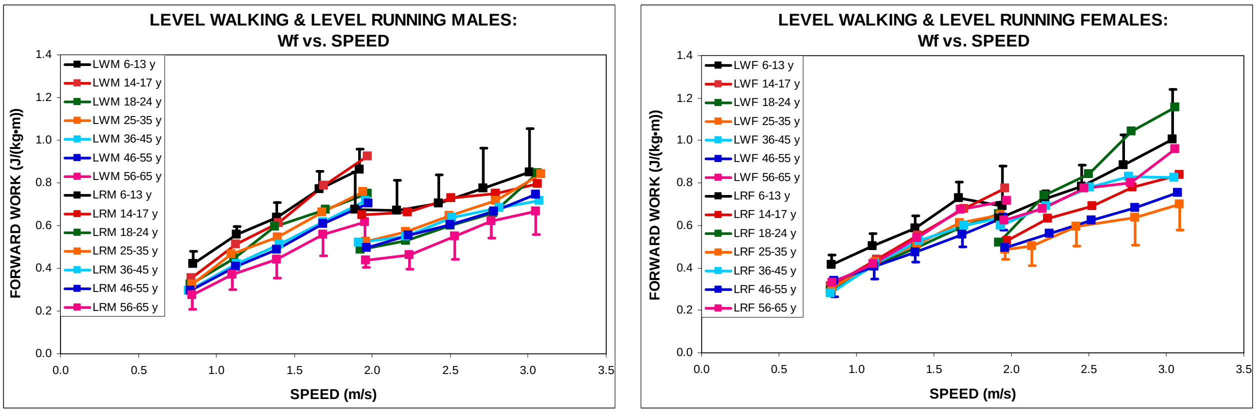


Figure 10.28. Forward work as a function of speed in level walking and running, males (on the left) and females (on the right).

All our results concur with literature data (both for W_v and W_f; Cavagna et al., 1983: Figure 10.29, left graph; Willems et al., 1995: Figure 10.29, right graph; Lejeune et al., 1998; Griffin et al., 1999; Cavagna et al., 2005).

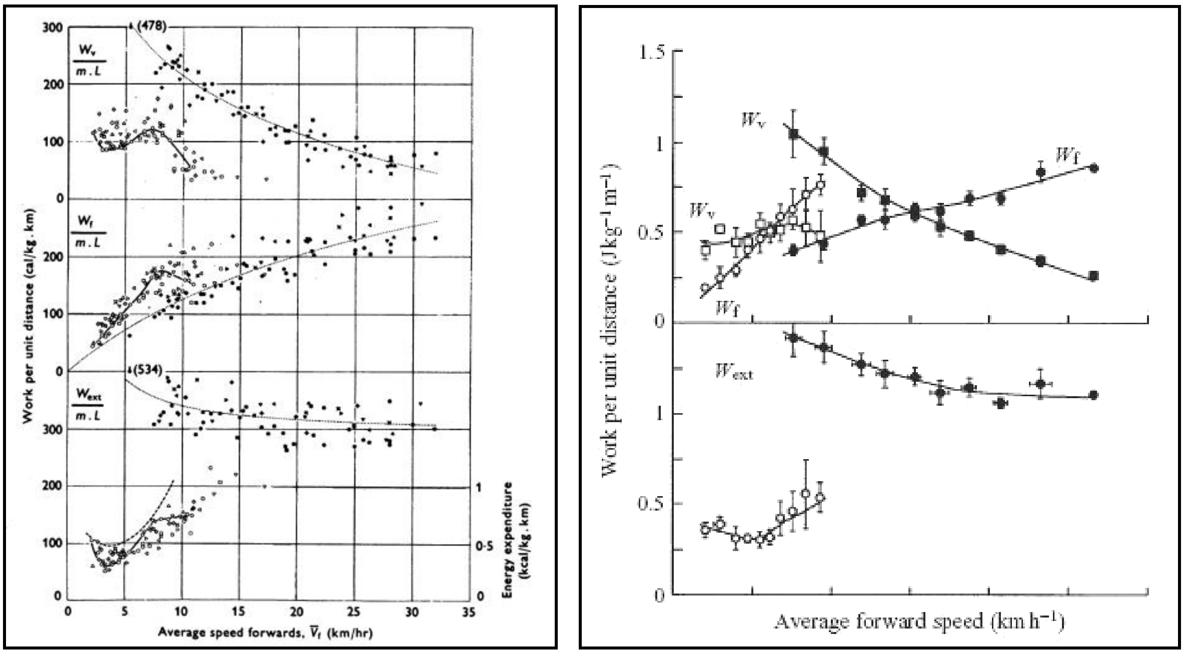


Figure 10.29. W_f and W_v as a function of speed, in Cavagna et al. (1983) (on the left) and in Willems et al. (1995) (on the right).

In conclusion, if W_v and W_f are analysed as a function of gradient, our results show that:

- in gradient walking (Figure 10.30), W_v is an upward linear function of gradient above -5%. In order to fully understand this pattern, see par. 6.3.3 below. This model is similar both in males and females;

- in gradient running (Figure 10.30), W_v is an upward linear function of gradient. In order to fully understand this pattern, see par. 6.5.3 below. This model is similar both in males and females;
- in both gradient walking and running (Figure 10.31), W_f does not significantly change with gradient.

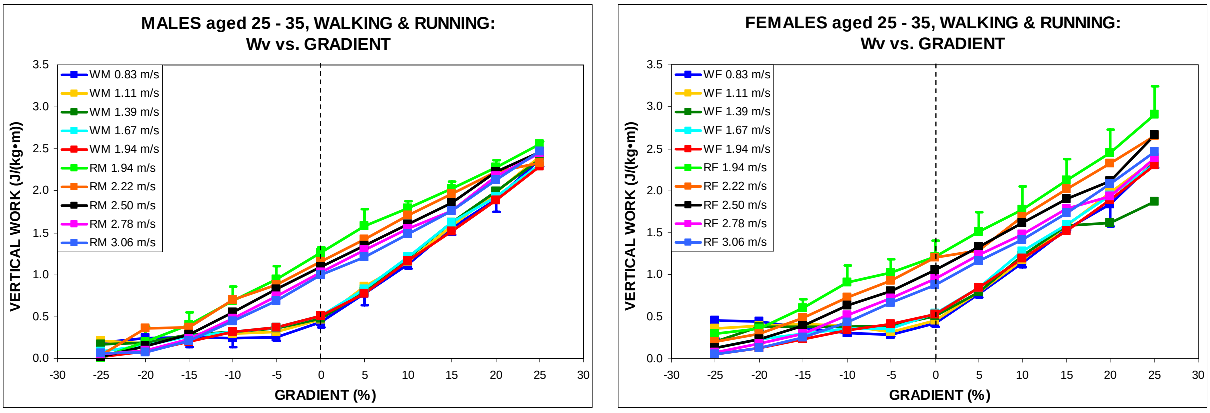


Figure 10.30. Vertical work as a function of gradient in walking and running, males (on the left) and females (on the right).

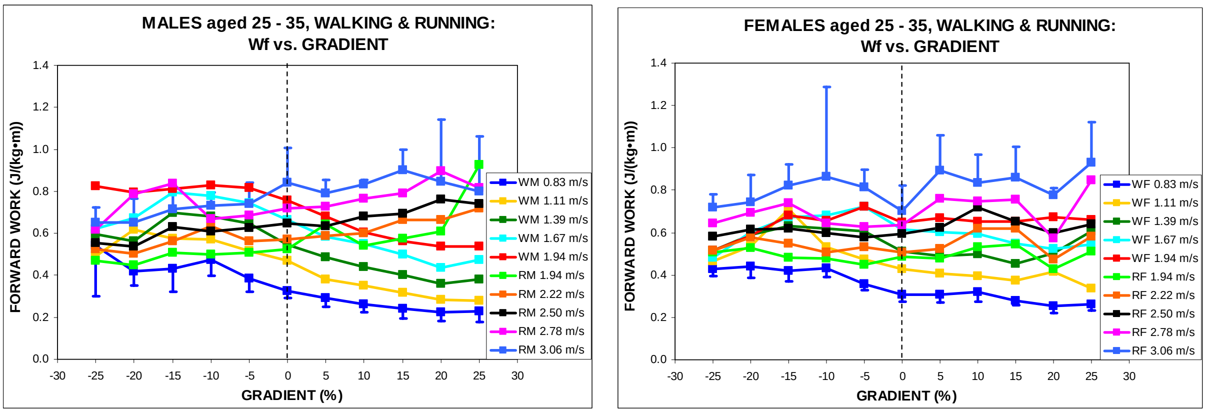


Figure 10.31. Forward work as a function of gradient in walking and running, males (on the left) and females (on the right).

6.3. Mechanical external work in walking

During each single step, in walking, the kinetic and potential energy are exchanged such that the total external work required to lift and accelerate the body's centre of mass is less than the sum of positive increments in potential and kinetic energy. Moreover, W_{ext} is always less than the sum of W_v and W_f since mechanical energy is recovered within the stride (Griffin et al., 1999).

According to literature (i.e. the paradox of mechanical work), in our experiments, in level locomotion, we have used the conventional approach of considering only positive increments in mechanical energy and neglecting negative work. Indeed, in level locomotion at a steady speed negative work is equal in magnitude but opposite in sign to positive work. Inclusion of negative work would result in a net work of zero which although mechanically correct is biologically meaningless (Mian et al., 2006). The conventional approach avoids this is called ‘zero work paradox’. However, in gradient locomotion, we did measure the W_{ext}^{-} corresponding to a decrease in total mechanical energy (Minetti et al., 1993; Saibene et al., 2003).

Particularly, mechanical external work is normalised to body mass and unit distance (J/(kg·m)).

6.3.1. Mechanical external work as a function of age

Precisely, our results show that:

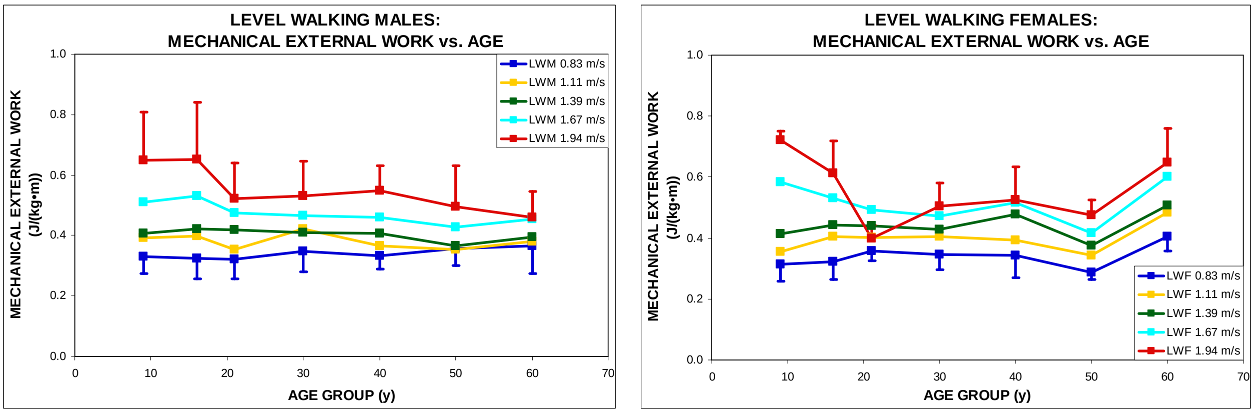


Figure 10.32. Mechanical external work as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), there is no significant variability among all age groups (Mian et al., 2006; Hernandez et al., 2009);
- however, in females (right graph), at the highest speed (1.94 m/s), young children aged 6 to 17 (0.721 ± 0.029 J/(kg·m), at all the investigated speeds) present the most specific pattern which differs from the other age groups (0.424 ± 0.069 J/(kg·m), p<0.001);
- at the other speeds, there is no significant variability among all age groups (0.555 ± 0.075 J/(kg·m), independently of age and speed);
- all our results concur with literature data (Mian et al., 2006; Ortega et al., 2007; Hernandez et al., 2009).

Specific results of the statistical analysis (with relevance) are shown in par. 4.1, Table 10.11 (see Appendix 10.1).

6.3.2. Mechanical external work as a function of speed

W_{ext} is an upward linear function of speed ($p<0.001$ both in males and females; Figure 10.33; Cavagna et al., 1983: Figure 10.34; Ardigò, 1992; Willems et al., 1995; Griffin et al., 1999; Saibene et al., 2003; Hallemans et al., 2004; Cavagna et al., 2005; Mian et al., 2006; Adamczyk et al., 2009; van Engelen et al., 2009).

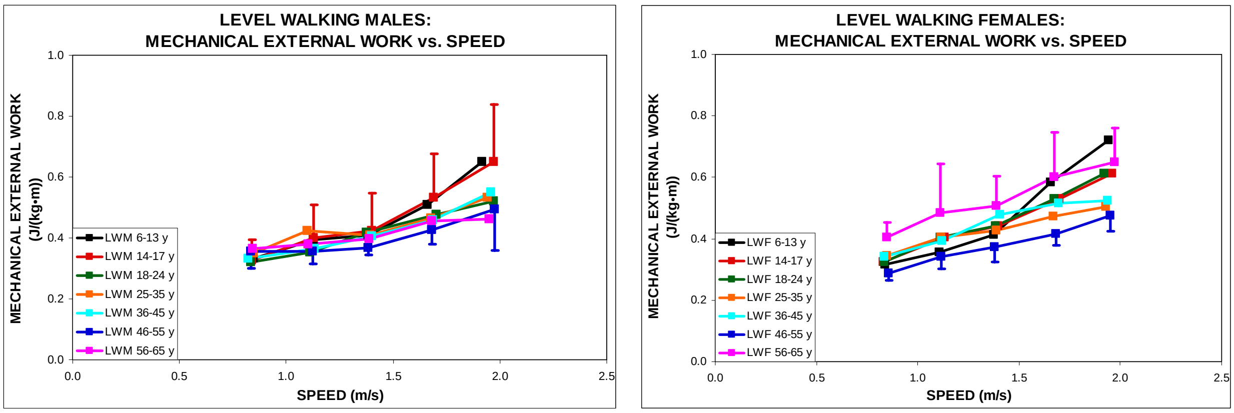


Figure 10.33. Mechanical external work as a function of speed in level walking, males (on the left) and females (on the right).

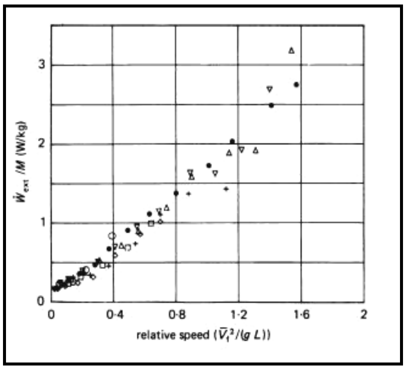


Figure 10.34. W_{ext} as a function of walking speed, in Cavagna et al. (1983).

Specific results of the statistical analysis (with relevance) are shown in par. 4.2, Table 10.12 (see Appendix 10.1).

6.3.3. Mechanical external work as a function of gradient

For males and females aged 25 to 35, we analysed mechanical external work in relation to gradient, too (Figure 10.35).

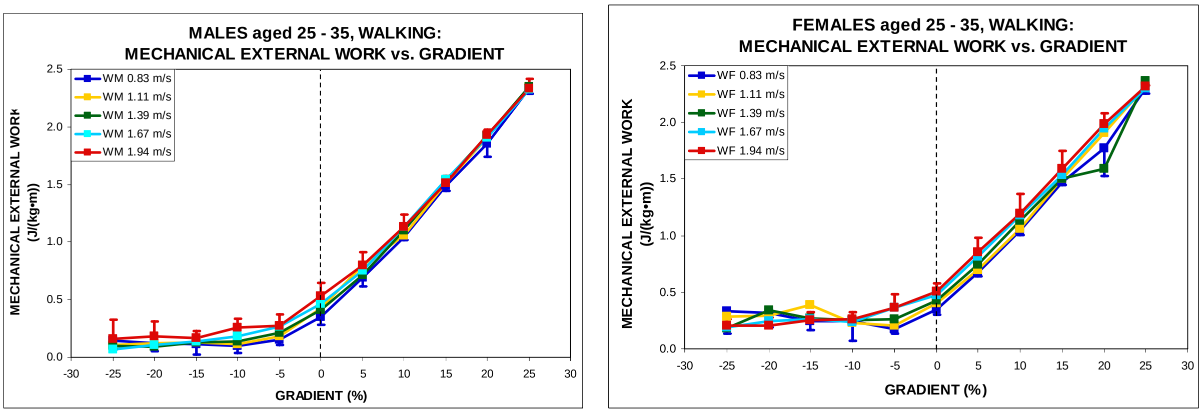


Figure 10.35. Mechanical external work as a function of gradient in walking, males (on the left) and females (on the right).

- W_{ext} is an upward linear function of gradient;
- particularly, it does not significantly change from -25 to -5% (from 0.116 ± 0.092 to 0.216 ± 0.050 J/(kg·m), at all the investigated speeds, in males; and from 0.239 ± 0.101 to 0.271 ± 0.086 J/(kg·m), in females);
- furthermore, above this gradient, W_{ext} increases linearly up to maximum gradient ($p<0.001$; Cavagna et al., 1976; 1977);
- this pattern occurs both in males (left graph) and females (right graph), at all speeds;
- on average, female values are slightly higher than male values;
- all our results concur with literature data (Minetti et al., 1993: Figure 10.36).

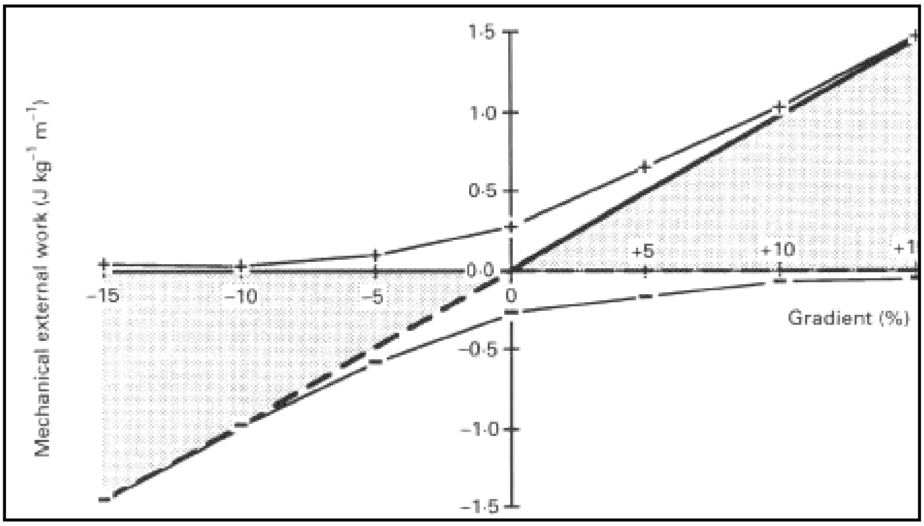


Figure 10.36. W_{ext} as a function of gradient, in Minetti et al. (1993).

Specific results of the statistical analysis (with relevance) are shown in par. 4.3, Table 10.13 (see Appendix 10.1).

6.4. Mechanical external work in running

It has been demonstrated that the mechanical external work done per unit distance in running appeared to be independent of speed (Cavagna et al., 1964). Particularly, a little doubt that it increases with speed remained (Cavagna et al., 1976). In fact, it has been shown that, in running, $W_{int} < W_{ext}$ up to 5.56 m/s (= 20 km/h), whereas at higher speeds $W_{int} > W_{ext}$.

More recently, it has been shown that, for adults running at different imposed step frequencies, a trade-off exists because the external power W_{ext} decreases with increasing step frequency (Cavagna et al., 1991). Furthermore, in adults running at a given speed with different step frequencies, W_{ext} decreases (Schepens et al., 2001). This decrease in W_{ext} is mainly due to a reduction in W_v and, to a much lesser extent, to a reduction in W_f (Cavagna et al., 1997).

6.4.1. Mechanical external work as a function of age

W_{ext} is higher in running (it ranges from 1.207 ± 0.055 to 1.758 ± 0.024 J/(kg·m), independently of age and speed, in males; and from 1.318 ± 0.032 to 1.787 ± 0.012 J/(kg·m), in females) than in walking (from 0.321 ± 0.065 to 0.649 ± 0.158 J/(kg·m), in males; and from 0.287 ± 0.024 to 0.721 ± 0.029 J/(kg·m), in females; Cavagna et al., 1976).

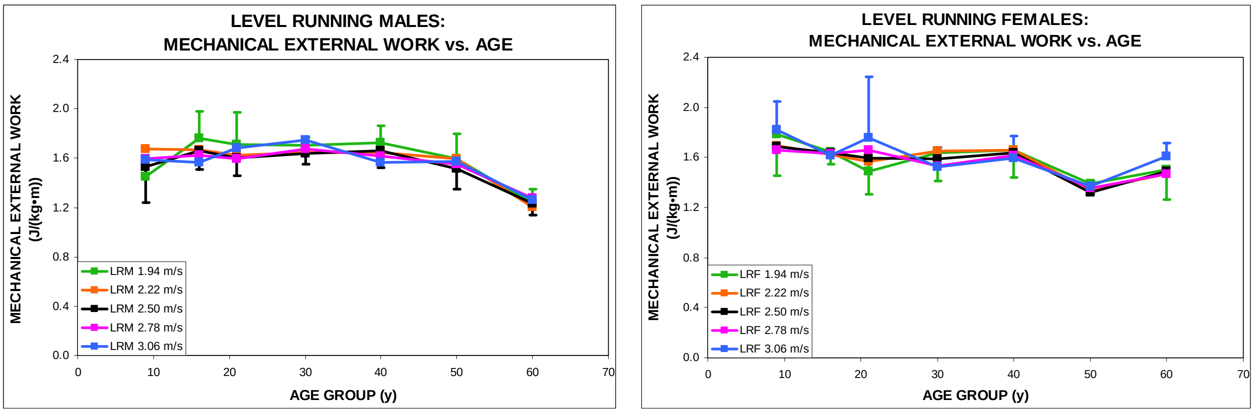


Figure 10.37. Mechanical external work as a function of age in level running, males (on the left) and females (on the right).

In level running, our results show that:

- only at the highest speed (3.06 m/s), in males (left graph), W_{ext} little changes with age ($p<0.001$ from 25 to 35 years to 56 to 65 years; Cavagna et al., 2008b);
- however, there is no significance at the other speeds, in each age group;
- it is also important to underline that W_{ext} does not change with age in females (right graph).

Specific results of the statistical analysis (with relevance) are shown in par. 4.4, Table 10.14 (see Appendix 10.1).

6.4.2. Mechanical external work as a function of speed

There is no significant change in W_{ext} as a function of speed both in males (left graph) and females (right graph) (Luhtanen et al., 1978).

However, the qualitative analysis shows that W_{ext} is lower in elderly subjects (1.243 ± 0.094 J/(kg·m), at all the investigated speeds, in males aged 56 to 65; and 1.431 ± 0.142 J/(kg·m), in females aged 46 to 65; Cavagna, 2008b: Figure 10.39).

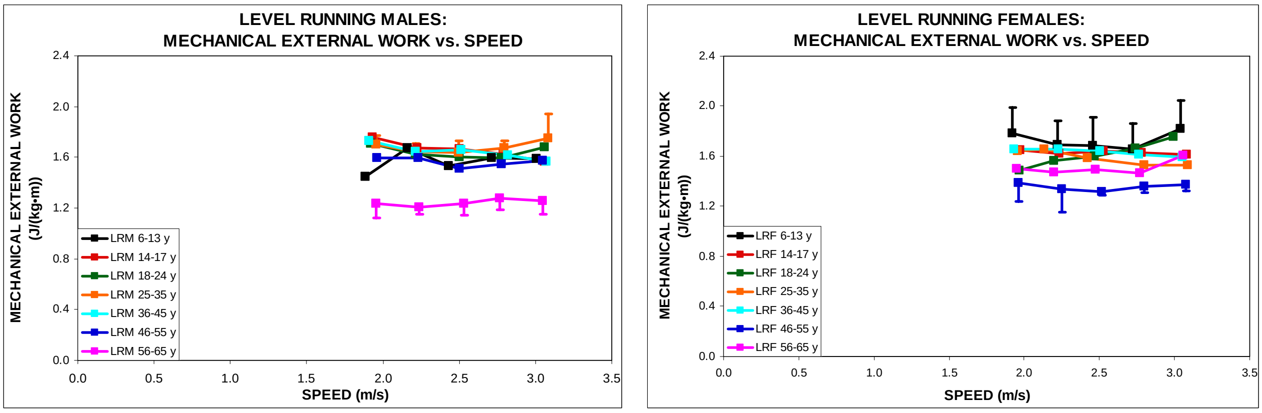


Figure 10.38. Mechanical external work as a function of speed in level running, males (on the left) and females (on the right).

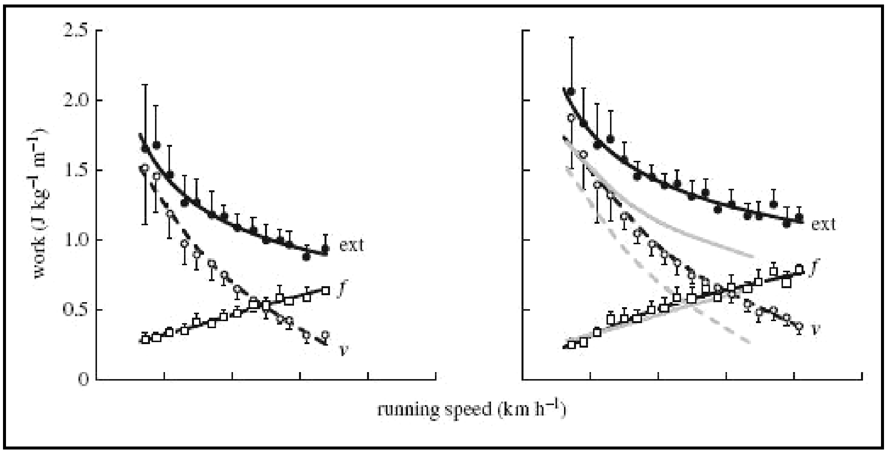


Figure 10.39. W_{ext} as a function of running speed, in Cavagna et al. (2008b). The external work is lower in the old subjects (73.6 ± 5.5 years; on the left) than in the young subjects (20.8 ± 1.6 years; on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 4.5, Table 10.15 (see Appendix 10.1).

6.4.3. Mechanical external work as a function of gradient

For males and females aged 25 to 35, we analysed the mechanical external work in relation to gradient, too (Figure 10.40).

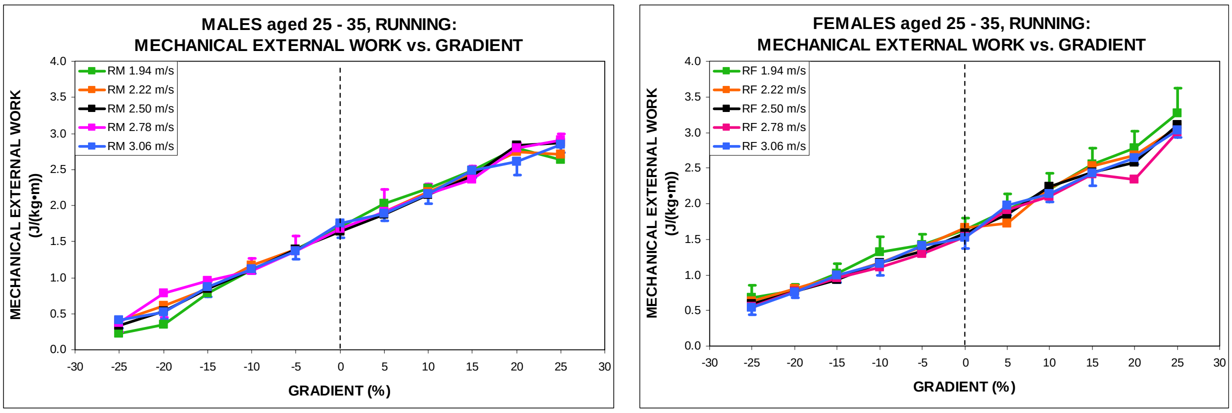


Figure 10.40. Mechanical external work as a function of gradient in running, males (on the left) and females (on the right).

- W_{ext} is an upward linear function of gradient ($p<0.001$; Cavagna et al., 1976; Flynn et al., 1993; Roberts et al., 2005);
- this pattern is similar both in males (left graph) and females (right graph);
- for instance, the qualitative analysis shows that, in males, W_{ext} increases with gradient up to 20% (from 0.348 ± 0.100 to 2.754 ± 0.085 J/(kg·m), at all the investigated speeds); above this gradient, it does not change;
- furthermore, this pattern is similar at each speed;
- all our results concur with literature data (Flynn et al., 1993; Minetti et al., 1993).

Specific results of the statistical analysis (with relevance) are shown in par. 4.6, Table 10.16 (see Appendix 10.1). Statistical analysis was not applied to female data, because of some discarded tests (see chapters 5, par. 1.1.2 and 6, par. 2.2).

7. ENERGY RECOVERY PERCENTAGE

7.1. Introduction

The amount of transfer between gravitational potential energy (PE) and kinetic energy (KE) is quantified by the energy recovery percentage (R) (Griffin et al., 1999; Minetti et al., 2001; Hallemans et al., 2004; Ivanenko et al., 2004; Mian et al., 2006; Usherwood et al., 2008; Mahaudens et al., 2008; 2009; van de Walle et al., 2009).

Introduced by Cavagna et al. (1976) to account the ability of a moving system to save energy by behaving like a pendulum (Saibene et al., 2003), it can be calculated, both from motion analysis and platform dynamometry, as:

$$R = \frac{\left(|W_f| + |W_v|\right) - W_{ext}}{|W_f| + |W_v|} \bullet 100 \quad \text{[Eq. 10.3]}$$

where W_f is the mechanical forward work over an integral number of strides (Mian et al., 2006; see par. 5.2 above); W_v is the mechanical vertical work over an integral number of strides (see par. 5.2 above); and W_{ext} is the mechanical external work.

In a frictionless pendulum (see also chapter 1, par. 4.3), the W_{ext} is nil and, as a consequence, R = 100%. In fact, all of the potential energy is 'recovered' as kinetic energy and *vice versa* during a cycle (Lejeune et al., 1998; Terrier et al., 2001; Hallemans et al., 2004; Ivanenko et al., 2004; Detrembleur et al., 2005; Ortega et al., 2005; Biewener, 2006; Kimura et al., 2009; Malatesta et al., 2009; Peyrot et al., 2009; Starke et al., 2009).

The recovery of mechanical energy through the pendulum-like mechanism of walking attains a maximum (about 65%) at intermediate walking speeds (*optimal speed*; Cavagna et al., 1976; Griffin et al., 1999; Cavagna et al., 2000; Alexander, 2004; Hallemans et al., 2004; Ivanenko et al., 2004; Ortega et al., 2005). Finally, it has been demonstrated that the most economical speed of walking is that at which the recovery is maximal (Saibene, 1990; Saibene et al., 2003).

Moreover, it has been found that recovery depends on a) stride length (Minetti et al., 1995) and b) walking speed (Cavagna et al., 1976); and it changes with age and body size in children, attaining a maximum at lower speeds in the younger subjects (Cavagna et al., 1983; Saibene, 1990; Hallemans et al., 2004). Furthermore, it varies between healthy and pathological subjects (Malatesta et al., 2009; van de Walle et al., 2009).

In the children, as in adults, the external work done per unit distance reaches a minimum near the speed at which the energy recovery is at a maximum (Cavagna et al., 1983). In addition:

- the minimum of the external work done per unit distance and the maximum transfer between the potential and kinetic energy are attained at the optimal speed which is the smaller the younger the subject;
- the amount of exchange is greatest at moderate walking speed (≈ 1.5 m/s for adults), because the energy fluctuations are nearly equal in magnitude and are 180° out of phase (Griffin et al., 1999);
- above the optimal speed the percentage energy recovery decreases, and the weight specific external work done per unit distance increases more steeply the smallest the age.

For this important biomechanical variable, see also chapter 11.

7.2. Results of our experiments

7.2.1. Energy recovery percentage as a function of age

Precisely, our results show that:

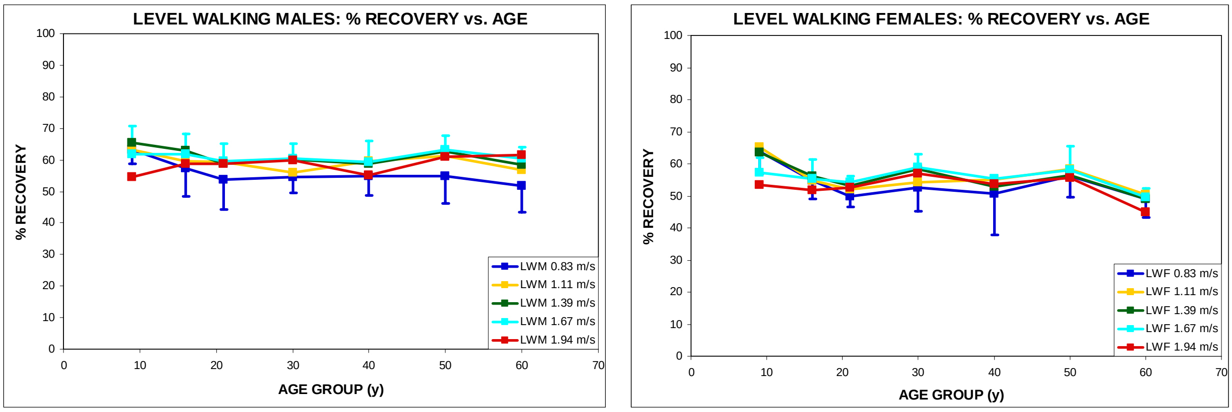


Figure 10.41. Energy recovery percentage as a function of age in level waking, males (on the left) and females (on the right).

- in males (left graph), there are no significant differences in all age groups;
- in females (right graph), energy recovery is slightly greater in young children (aged 6 to 13, $p<0.05$). This patterns occurs at 1.39 (63.538 ± 3.555%) and 1.94 m/s (53.330 ± 10.097%);
- at the other speeds, there is no significance in R as a function of age.

Specific results of the statistical analysis (with relevance) are shown in par. 5.1, Table 10.17 (see Appendix 10.1).

7.2.2. Energy recovery percentage as a function of speed

The maximal recovery speed increases the older the subject, or the 'optimal speed' of walking is the smaller the younger the subject (Ralston, 1958; Cavagna et al., 1976: Figure 10.43, left graph; 1983; Willems et al., 1995: Figure 10.43, right graph; Lejeune et al., 1998; Griffin et al., 1999; Hallemans et al., 2004; Mian et al., 2006; Kimura et al., 2009).

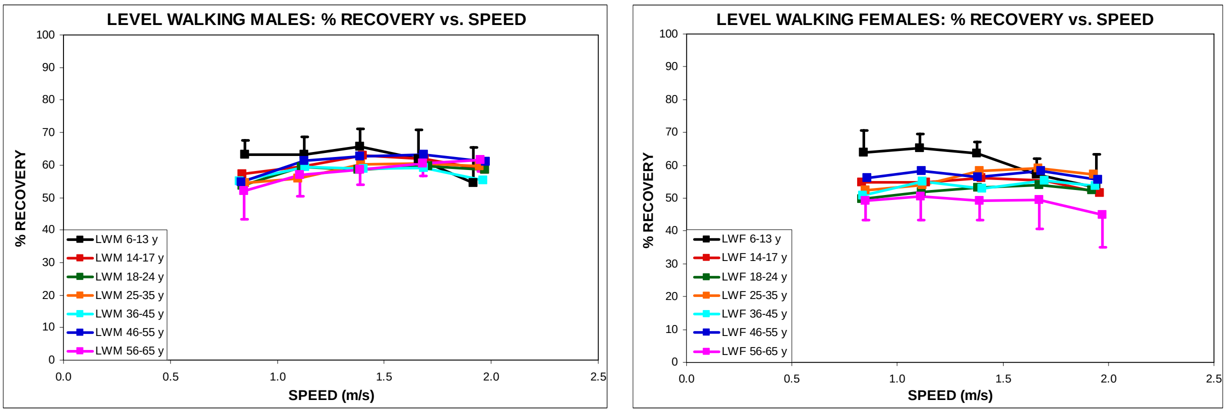


Figure 10.42. Energy recovery percentage as a function of speed in level walking, males (on the left) and females (on the right).

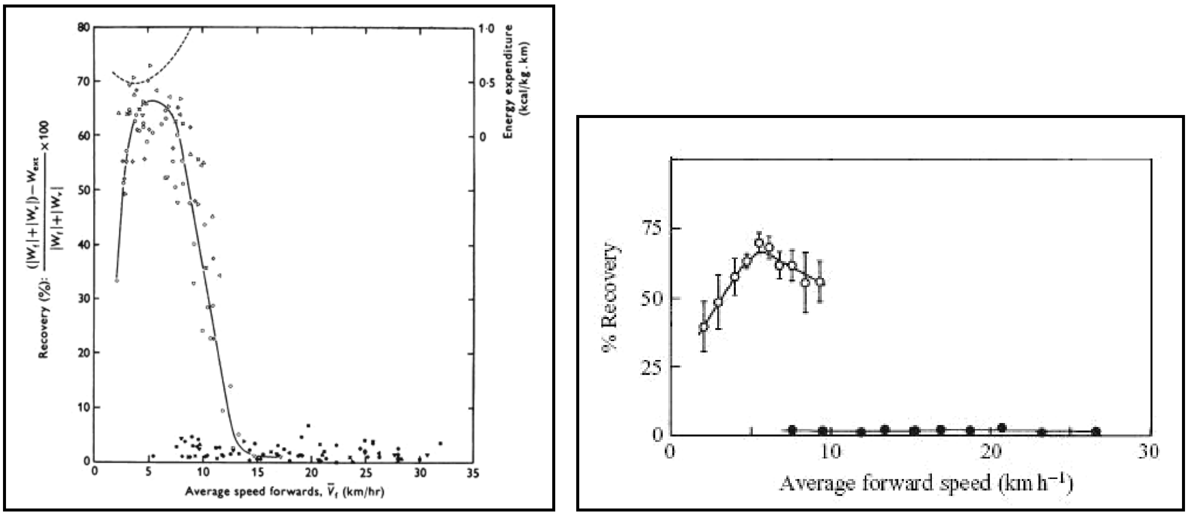


Figure 10.43. R as a function of speed, in Cavagna et al. (1976) (on the left) and in Willems et al. (1995) (on the right).

Particularly, our results show that:

- this optimal speed ranges from ≈ 1.11 to ≈ 1.39 m/s in young subjects aged 6 to 13, and it is higher than ≈ 1.39 m/s in young and elderly adults aged 14 to 65;

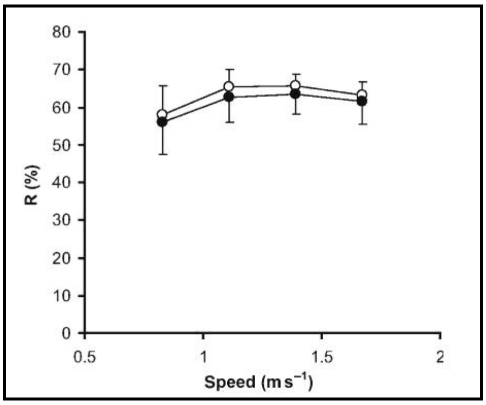


Figure 10.44. R as a function of speed, in Mian et al. (2006).
Open circles refer to young subjects; solid circles to old subjects.

- above this ‘optimal speed’, the energy recovery decreases (Cavagna et al., 1976, 1983; Willems et al., 1995; Lejeune et al., 1998; Hallemans et al., 2004);
- this is more evident in young children (both in males and females, aged 6 to 13), and in elderly females (aged 56 to 65).

Specific results of the statistical analysis (with relevance) are shown in par. 5.2, Table 10.18 (see Appendix 10.1).

In level running (Figure 10.45), our results concur with literature data (Cavagna et al., 1976; Willems et al., 1995; Lejeune et al., 1998; Alexander, 2004; Biewener, 2006).

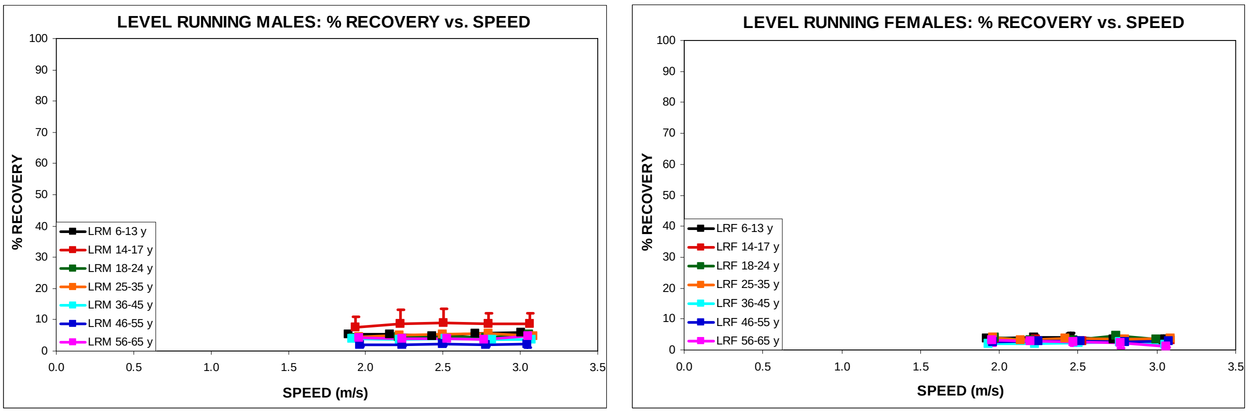


Figure 10.45. Energy recovery percentage as a function of speed in level running, males (on the left) and females (on the right).

7.2.3. Energy recovery percentage as a function of gradient

For males and females aged 25 to 35, we analysed the energy recovery percentage in relation to gradient, too (Figure 10.46).

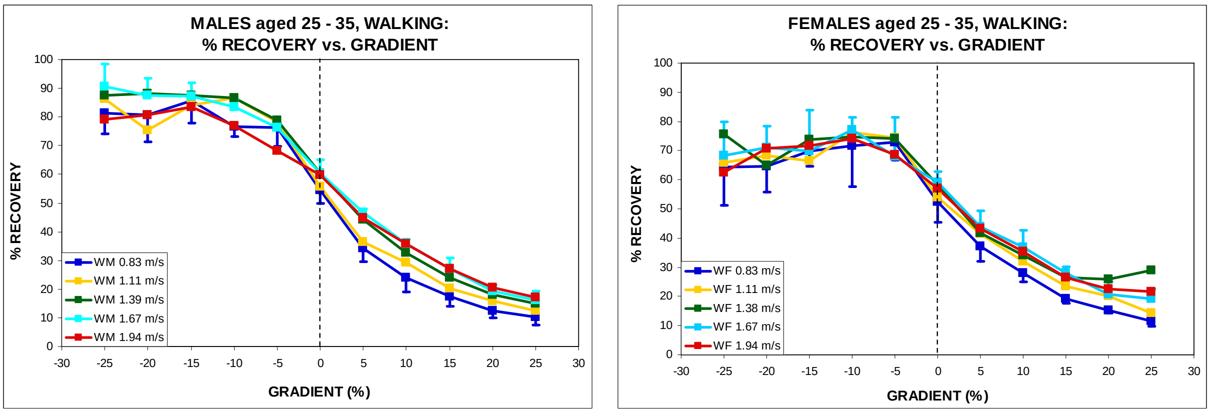


Figure 10.46. Energy recovery percentage as a function of gradient in walking, males (on the left) and females (on the right).

Precisely, in males (left graph), our results show that:

- energy recovery is a downward linear function of gradient (Minetti et al., 1993);
- to be more specific, from -25 to -10%, R does not significant change with gradient (from 84.973 ± 7.906 to 82.122 ± 4.038%, at all the investigated speeds);
- moreover, from -10% to the level condition, R presents a sigmoidal pattern (from 82.122 ± 4.038 to 58.182 ± 5.669%);
- finally, it decreases with gradient from the level condition to the maximum up gradient ($p<0.001$).

In females (right graph):

- energy recovery is a downward linear function of gradient;

- precisely, from -25 to -10%, R does not significant change with gradient (from 67.359 ± 11.869 to 74.783 ± 8.063%, at all the investigated speeds);
- moreover, from -10% to the level condition, R presents a sigmoidal pattern (from 74.783 ± 8.063 to 56.203 ± 5.265%);
- finally, it decreases with gradient from the level condition to the maximum up gradient ($p<0.001$);
- moreover, the qualitative analysis shows that, in females, from 20 to 25%, energy recovery little increases (from 20.943 ± 3.345 to 19.051 ± 6.106%, at all the investigated speeds. This pattern is probably related to their higher vertical displacement of the BCOM at these slopes).

Specific results of the statistical analysis (with relevance) are shown in par. 5.3, Table 10.19 (see Appendix 10.1).

On average, at the lowest gradients (from -25 to -10%), energy recovery is little higher in males (83.788 ± 6.241%, independently of speed and gradient) than in females (70.083 ± 9.100%. This is probably due a lowest average optimal walking speed in females compared to males, according to literature data). Finally, it is important to highlight our inverse proportionality relationship between W_{ext} and R (Cavagna et al., 1983).

8. MECHANICAL INTERNAL WORK

8.1. Mechanical internal work definition

The work necessary to accelerate the limbs with respect to the BCOM during locomotion is known as mechanical internal work (W_{int}) (Cavagna et al., 1976; Aleshinsky, 1986b; 1986c; Steudel 1990a; 1990b; Lejeune et al., 1998; Cerretelli, 2001; Minetti et al., 2001a; 2001b; Kautz et al., 2002; Bastien et al., 2003; Hallemans et al., 2004; Minetti, 2004; Mian et al., 2006; Minetti et al., 2006; van de Hecke et al., 2007; Mahaudens et al., 2008; Sasaki et al., 2008; Genin et al., 2009; Mahaudens et al., 2009). In contrast to W_{ext}, the main interest in W_{int} resides in the capability to consider the acceleration of body segments whose movements do not directly result in a change of the BCOM position (Minetti et al., 1998; Kautz et al., 2002).

The internal work involves a new biomechanical analysis that takes into account potential and kinetic energy components, all exchanges of energy within and between segments, and positive and negative work done by the muscles (Winter, 1979). This concept of internal work comes from the König theorem of physics (Cavagna and Kaneko, 1977) stating that '*in a linked multi-segment system, the total kinetic energy can be partitioned in the one of the BCOM with respect to the environment and the one of single segments with respect of the BCOM*'. The second component is then incorporated into the mechanical internal work. By summing the kinetic energy curves of single segments in a way which allows energy transfer only among within-limb segments, and by summing all the energy increases in the resulting curves, W_{int} can be calculated.

W_{int} proved to be useful in comparative and intra-species analysis of the mechanical relationship during locomotion in different conditions (Cavagna et al., 1997), gait (Minetti et al., 1994), gradient (Minetti et al., 1993) and stride frequency (Cavagna et al., 1986; 1991). It has been demonstrated that:

- W_{int} increases approximately as the square of the speed of walking and running (Cavagna et al., 1976);
- W_{int} is constant at each speed regardless of gradient. This is partly explained by a slightly decrease in stride frequency at increasing gradient. W_{int} constancy implies that it has no rule in determining the optimum gradient (Minetti et al., 1993);
- at any given speed, W_{int} increases with step frequency, also independent of age and body dimensions (Cavagna et al., 1986; Schepens et al., 2001);
- a higher step frequency involves a greater internal work to reset the limbs at each step (Cavagna et al., 2008b).

For other information concerning this important complex biomechanical variable, see also chapter 12.

8.2. Results of our experiments

As previously discussed in external work (see par. 6.3 above), according to literature (i.e. the paradox of mechanical work), in our experiments, in level locomotion, we calculated only the W_{int}^{+}. This is because of the nil value of W_{int} when walking and running occur at the level gradient with a constant speed. However, in gradient locomotion, we did measure the W_{int}^{-}, as well (Minetti et al., 1993; Saibene et al., 2003).

Particularly, mechanical internal work is normalised to body mass and unit distance (J/(kg·m)).

8.2.1. Mechanical internal work as a function of age

8.2.1.1. Mechanical internal work in level walking

In level walking, in males (left graph) and females (right graph), W_{int} does not significantly change with age. These results partially concur with data presented in Mian et al. (2006).

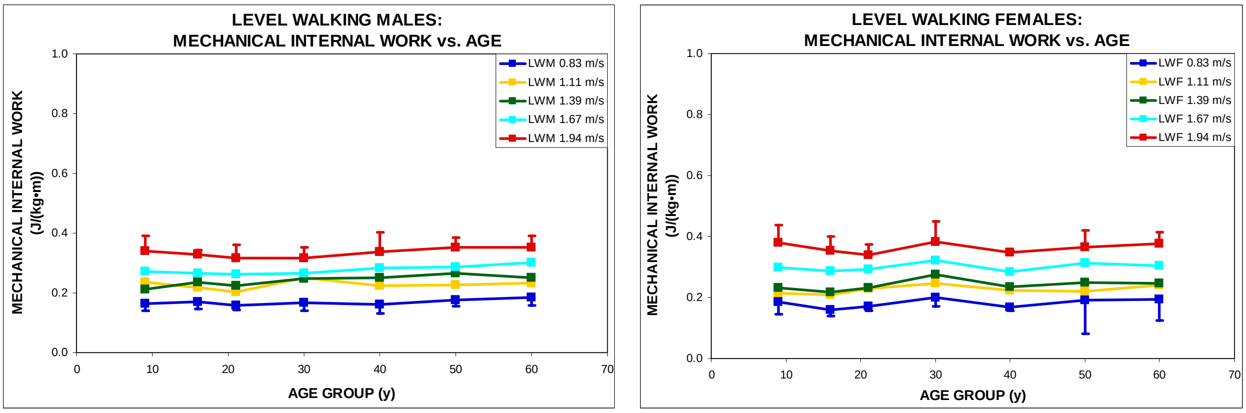


Figure 10.47. Mechanical internal work as a function of age in level walking, males (on the left) and females (on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 6.1, Table 10.20 (see Appendix 10.1).

8.2.1.2. Mechanical internal work in level running

In level running, in males (left graph) and females (right graph), W_{int} does not significantly change with age. As in walking, these results partially concur with data in Mian et al. (2006).

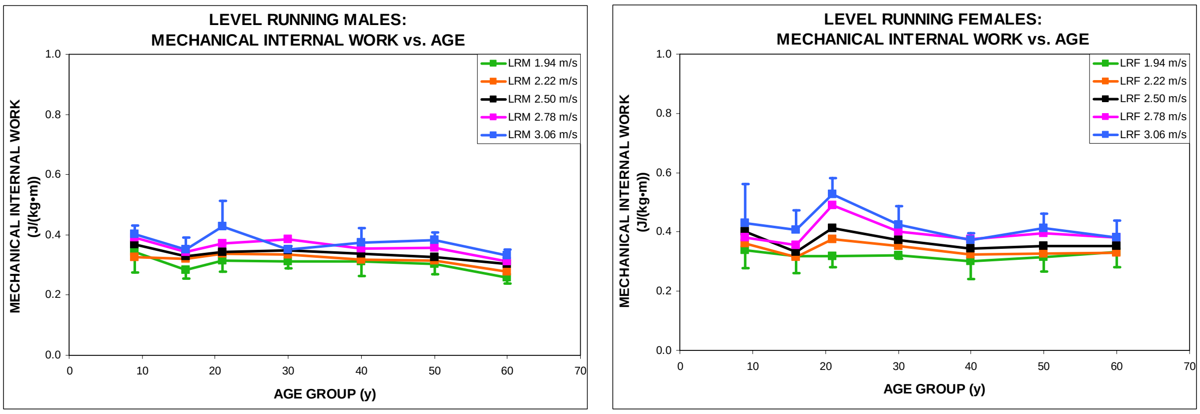


Figure 10.48. Mechanical internal work as a function of age in level running, males (on the left) and females (on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 6.2, Table 10.21 (see Appendix 10.1).

8.2.2. Mechanical internal work as a function of speed

In our results:

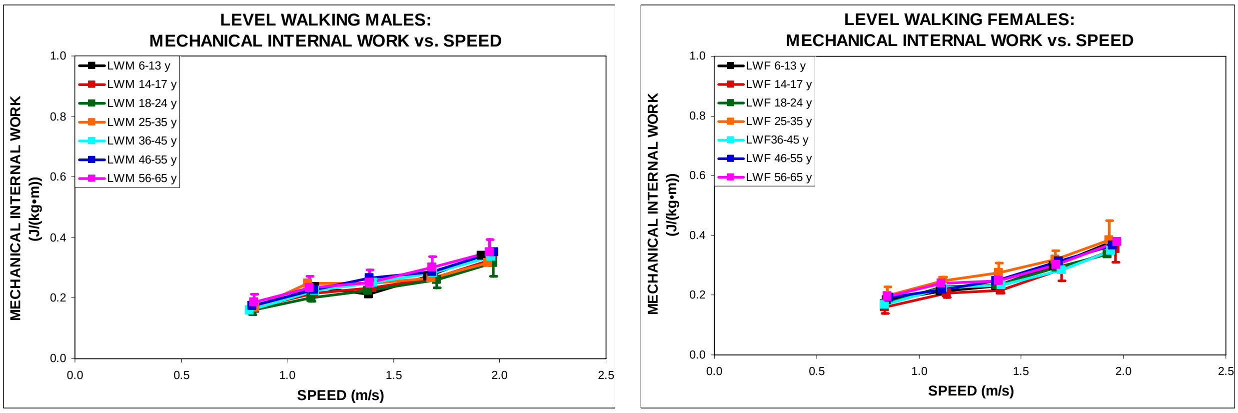


Figure 10.49. Mechanical internal work as a function of speed in level walking, males (on the left) and females (on the right).

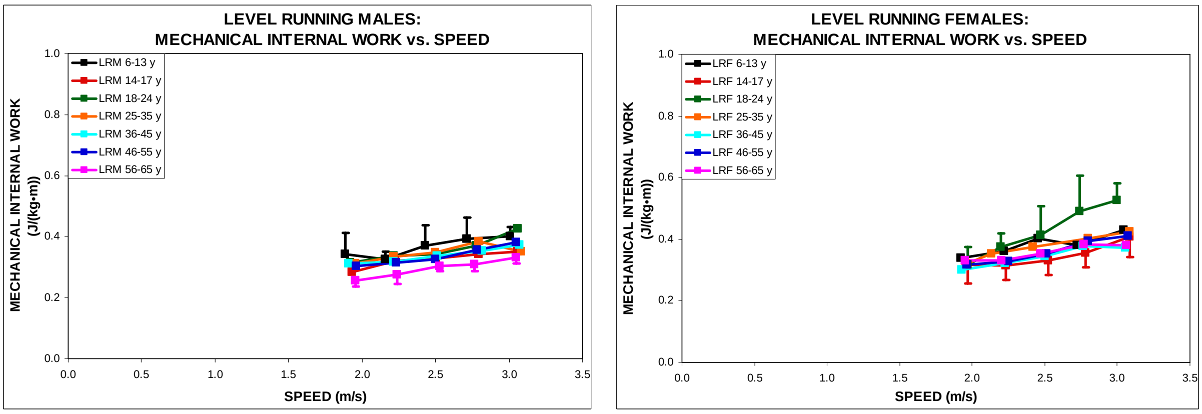


Figure 10.50. Mechanical internal work as a function of speed in level running, males (on the left) and females (on the right).

- W_{int} is little higher in running than in walking (Figure 10.50 *versus* 10.49; Cavagna et al., 1986; Willems et al., 1995; Lejeune et al., 1998; Minetti, 1998; Minetti et al., 2001b; Saibene et al., 2003). Particularly, in level walking, W_{int} ranges from 0.160 ± 0.029 to 0.352 ± 0.041 J/(kg·m), at all the investigated speeds, in males; and from 0.159 ± 0.019 to 0.384 ± 0.067 J/(kg·m), in females. In level running, it ranges from 0.257 ± 0.020 to 0.428 ± 0.086 J/(kg·m), in males; and from 0.315 ± 0.048 to 0.526 ± 0.154 J/(kg·m), in females;
- generally, our mechanical internal work values are smaller than those reported by Cavagna et al. (1977), Cavagna et al. (1991), Willems et al. (1995), Minetti (1998) and Schepens et al. (2001). In detail, our children (aged 6 to 13) present the greatest values at the level gradient: this is probably due to the effect of leg length on the effective step frequency (Schepens et al., 2001). Moreover, differently from what illustrated in Cavagna et al. (2008), our male elderly adults (aged 56 to 65) present the lowest internal work at level running. This finding seems a bit odd because the stride frequency and the duty factor of these

subjects are greater with respect to other age groups. However, this result could be ascribed both to a more stabilized movement of the upper limbs and a different moving pattern;

- furthermore, W_{int} is an upward linear function of speed ($p<0.001$ both in males and females; Cavagna et al., 1964; 1983; Minetti et al., 1993; Willems et al., 1995: Figure 10.51, left and middle graph; Lejeune et al., 1998; Minetti, 1998: Figure 10.51b, right graph; Minetti et al., 2001b; Schepens et al., 2001; Saibene et al., 2003; Hallemans et al., 2004; Cavagna et al., 2008b). This pattern is evident at each speed.

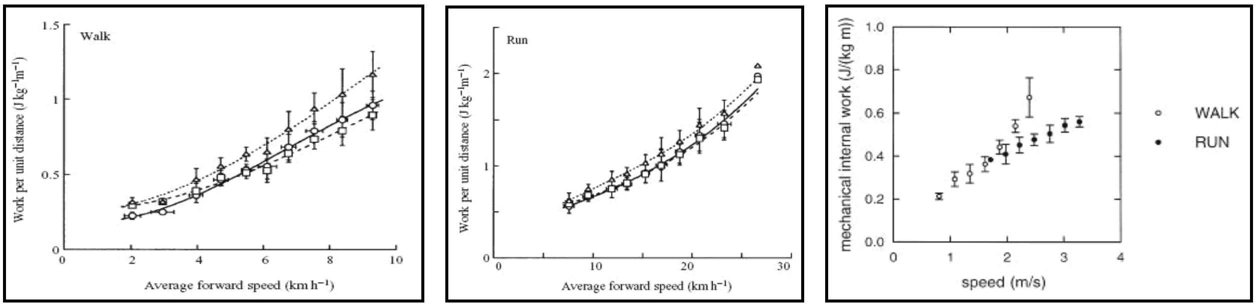


Figure 10.51. W_{int} as a function of speed, in Willems et al. (1995) (on the left and in the middle) and in Minetti (1998) (on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 6.3, Table 10.22 and 10.23 (see Appendix 10.1).

8.2.3. Mechanical internal work as a function of gradient

For males and females aged 25 to 35, we analysed the mechanical internal work in relation to gradient, too (Figure 10.52 and 10.54).

8.2.3.1. Mechanical internal work in gradient walking

Precisely, both in males (left graph) and females (right graph), our results show that:

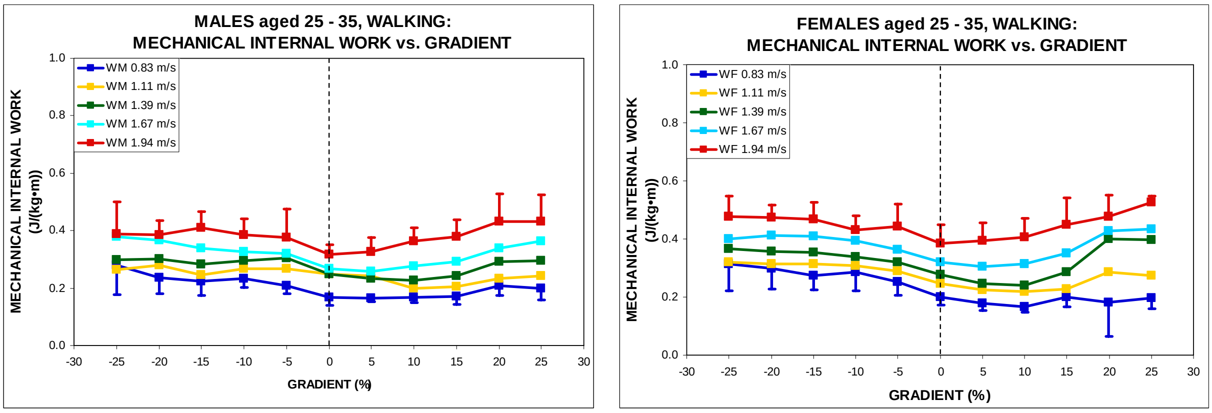


Figure 10.52. Mechanical internal work as a function of gradient in walking, males (on the left) and females (on the right).

- there is no significance in W_{int} as a function of gradient (Cavagna et al., 1976; 1997; Ardigò, 1992; Minetti et al., 1993: Figure 10.53; Saibene et al., 2003; Cavagna et al., 2008a). This pattern is quite similar at each speed;
- however at extreme slopes (both downhill and uphill), the qualitative analysis has underlined an increasing in W_{int}. Probably, this pattern is related to a major motor control that is necessary to sustain these slopes (Ardigò, 1992; Minetti et al., 1993).

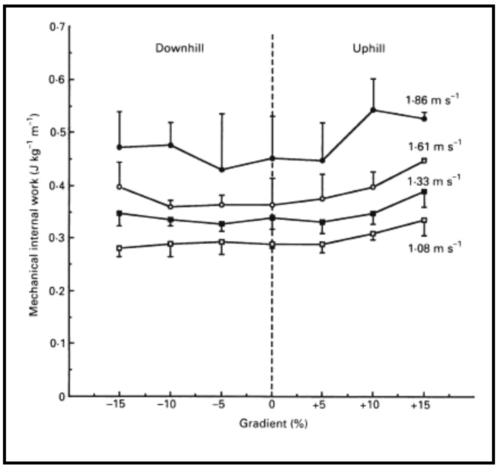


Figure 10.53. W_{int} as a function of gradient walking, in Minetti et al. (1993).

Specific results of the statistical analysis (with relevance) are shown in par. 6.5, Table 10.24 (see Appendix 10.1).

8.2.3.2. Mechanical internal work in gradient running

Precisely, in males (left graph), our results show that:

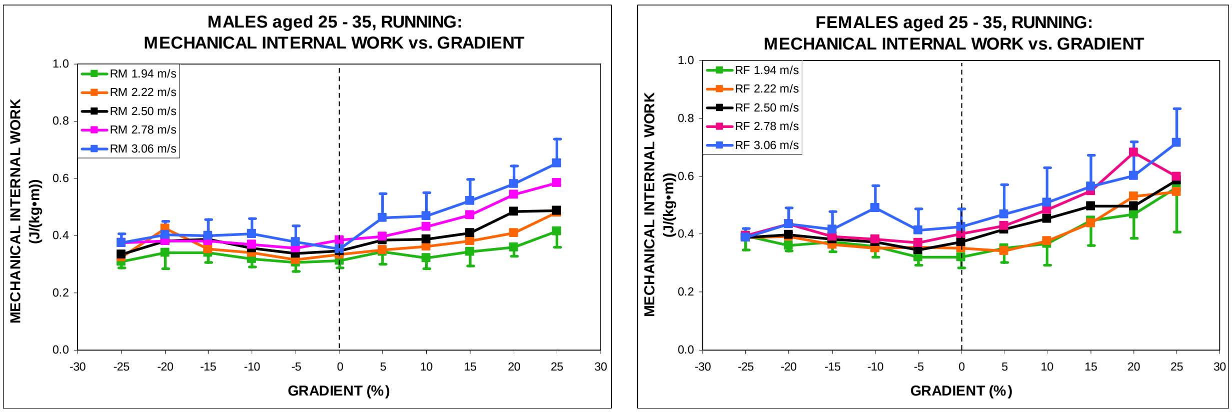


Figure 10.54. Mechanical internal work as a function of gradient in running,
males (on the left) and females (on the right).

- W_{int} does not significantly change with gradient (Cavagna et al., 1976; Ardigò, 1992; Minetti et al., 1994: Figure 10.55; Saibene et al., 2003);

- as shown in walking, however at extreme slopes (both downhill and uphill), the qualitative analysis has underlined an increasing in W_{int}. Probably, this pattern is related to a major motor control that is necessary to sustain these slopes (Ardigò, 1992; Minetti et al., 1993);

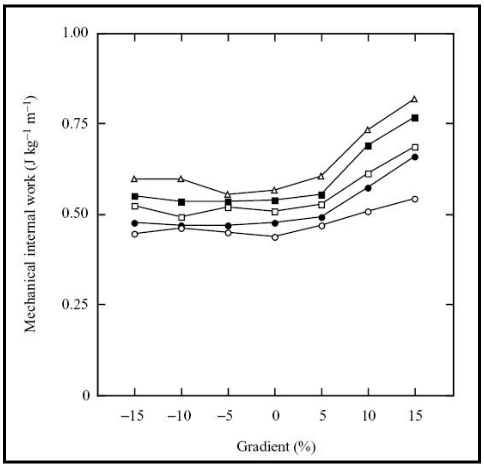


Figure10.55. W_{int} travelled at different speeds as a function of gradient running, in Minetti et al. (1994).

- this pattern is quite similar at each speed. The only exception is at the highest speed of 3.06 m/s (from 15%, 2.489 ± 0.103 J/(kg·m), to 25%, 2.848 ± 0.110 J/(kg·m), $p<0.05$).

Specific results of the statistical analysis (with relevance) are shown in par. 6.6, Table 10.25 (see Appendix 10.1).

Statistical analysis was not applied to female data, because of some discarded tests (see also chapters 5, par. 1.1.2 and 6, par. 2.2). However, in females (right graph), the qualitative analysis shows similar results to males.

9. MECHANICAL TOTAL WORK

9.1. Mechanical total work definition

Mechanical total work (W_{tot}) is calculated as the sum of W_{ext} and W_{int} (in absolute values: see par. 6 and 8 above), which are considered as two separate entities (Gordon et al., 1980; Aleshinsky, 1986a; 1986b; Minetti et al., 1993: Figure 10.56; Zatsiorsky, 1998; Sparrow, 2000; Minetti et al., 2001b; Donelan et al., 2002b; Kautz et al., 2002; Detrembleur et al., 2003; Saibene et al., 2003; Robertson et al., 2004; van de Hecke et al., 2007; Mahaudens et al., 2008; Neptune et al., 2008; Genin et al., 2009; Mahaudens et al., 2009):

$$W_{tot} = |W_{ext}| + |W_{int}| \quad \text{[Eq. 10.4]}$$

This resulted from an interpretation of König's theorem of mechanics stating that '*the overall KE of a linked multi-segmented system is the sum of the KE of the centre of mass of the system and* *those of the segments, calculated from their relative speeds from the centre of mass of the system*' (Burdett et al., 1983; Willems et al., 1995; Saibene et al., 2003). The first component has been incorporated in W_{ext}, while the second in W_{int}.

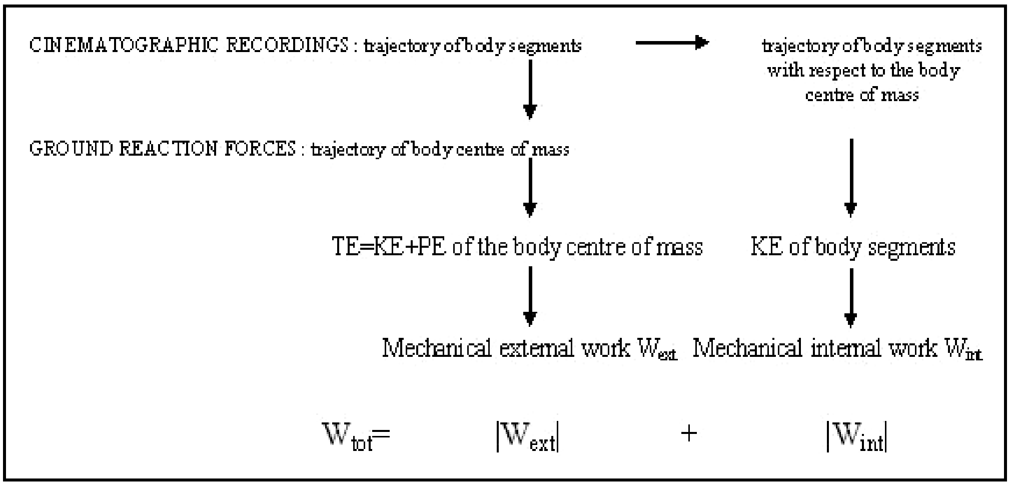


Figure 10.56. Method by Cavagna and Kaneko (1977), modified by Minetti et al. (1993).

As a consequence, a minimum value for W_{tot} is obtained by assuming complete energy transfer among the segments as well as between the BCOM and the segments (Willems et al., 1995).

The mechanical work performed when walking on a treadmill at different gradients (from -15 to +15%, step 5%) and speed (from 0.83 to 1.94 m/s, step 0.28 m/s) has been measured (Minetti et al., 1993). Importantly, it was observed that, while walking in a level manner, the positive and negative external work exerted were, as expected, equal but when the gradient of the treadmill changed, the fraction of positive work decreased in a sigmoidal pattern from almost 100% at 15% gradient to almost nil at -15% gradient, regardless of speed (as previously discussed in par. 6.3 and 8.2 for external and internal work, respectively). In the ranges between ±15%, the W_{ext} is not solely dependent on gradient, but also implies changes in kinetic energy (Saibene et al., 2003). In contrast, the W_{int} per unit distance, which increases linearly with the speed, is independent of the gradient (Saibene et al., 2003). In another important research, it has been verified that the muscle-tendon work of locomotion is most accurately measured when energy transfers are only included between segments of the same limbs, but not among the limbs or between the limbs and the centre of mass of the whole body (Willems et al., 1995). Finally, more recently, the mechanical work of both forward and backward walking on a treadmill at seven gradients (from 0 to +32%) has been investigated. With respect to forward locomotion, backward walking implies: a) the same mechanical internal work despite an increased stride frequency; b) higher mechanical external work within a gradient from 0 to 15%; c) lower energy recovery; and d) a decrease of the efficiency of locomotion particularly at 0% gradient (Minetti et al., 2001).

9.2. Results of our experiments

Particularly, mechanical total work is normalised to body mass and unit distance (J/kg·m).

9.2.1. Mechanical total work as a function of age

9.2.1.1. Mechanical total work in level walking

Precisely, our results show that:

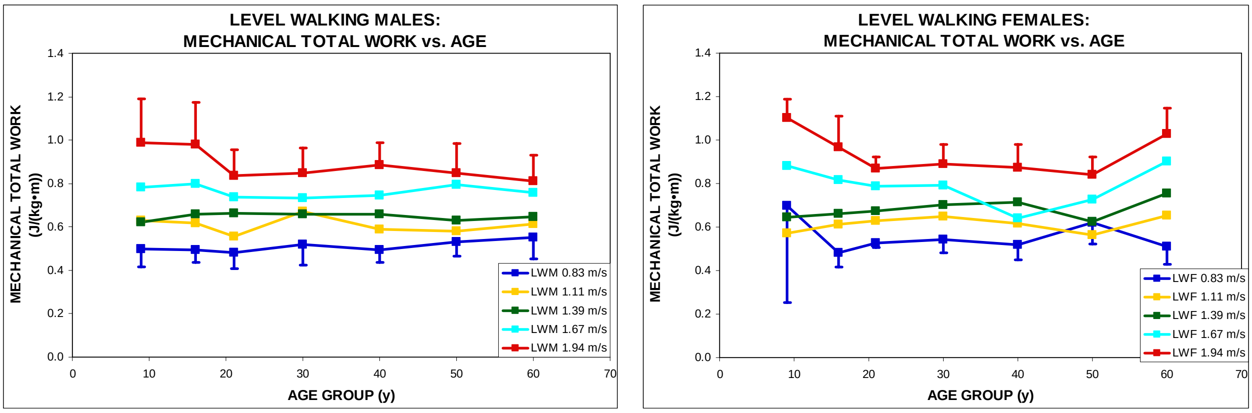


Figure 10.57. Mechanical total work as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), there is no significant change in W_{tot};
- in females (right graph), W_{tot} is higher in young subjects aged 6 to 13 (0.780 ± 0.172 J/(kg·m), at all the investigated speeds) and in elderly adults aged 56 to 65 (0.792 ± 0.100 J/(kg·m).

Specific results of the statistical analysis (with relevance) are shown in par. 7.1, Table 10.26 (see Appendix 10.1).

9.2.1.2. Mechanical total work in level running

Precisely, our results show that:

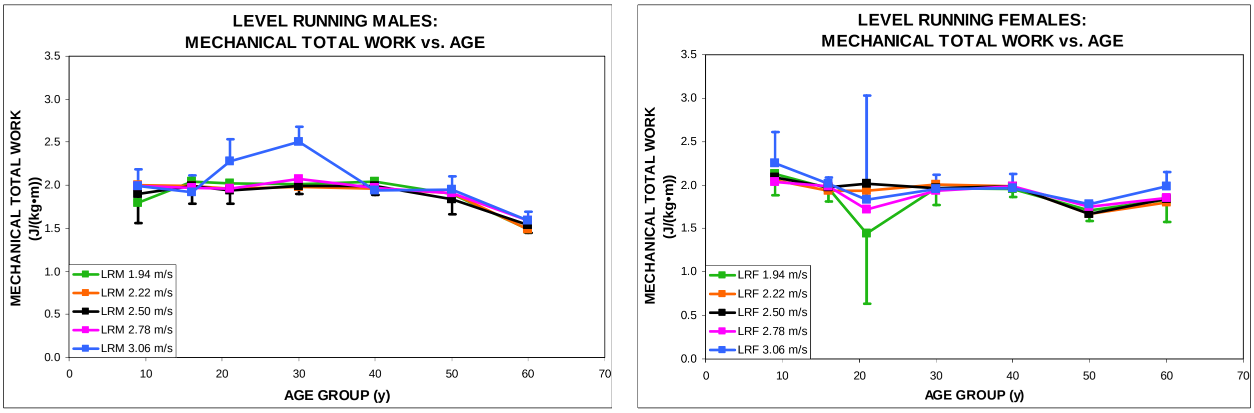


Figure 10.58. Mechanical total work as a function of age in level running,

males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), there is no significant change in W_{tot};
- this pattern is similar to what happens in W_{int} (Cavagna et al., 2008b).

Specific results of the statistical analysis (with relevance) are shown in par. 7.2, Table 10.27 (see Appendix 10.1).

9.2.2. Mechanical total work as a function of speed

In walking:

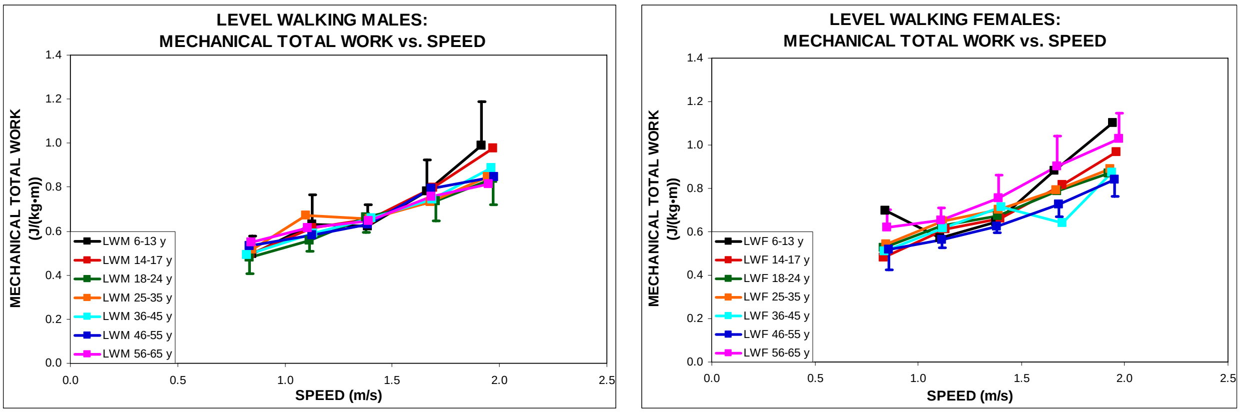


Figure 10.59. Mechanical total work as a function of speed in level walking, males (on the left) and females (on the right).

- there is a significance between W_{tot} and speed ($p<0.001$ both in males and females; Cavagna et al., 1964; 1976; Ardigò, 1992; Schepens et al., 1998; Hallemans et al., 2004);
- furthermore, in this case, W_{tot} is primarily dependent on the corresponding and similar pattern of W_{int} (see par. 7.2.2 above).

However, in running:

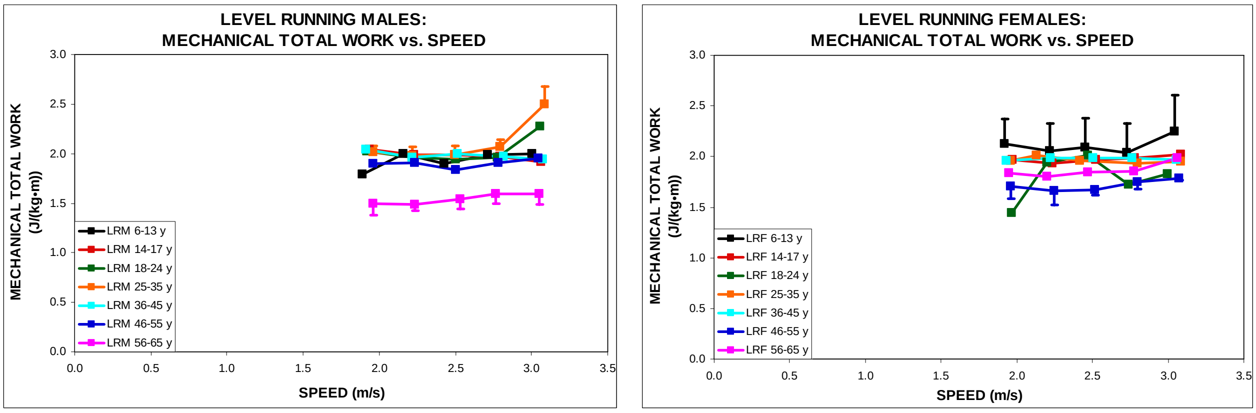


Figure 10.60. Mechanical total work as a function of speed in level running, males (on the left) and females (on the right).

- there is no significance between W_{tot} and speed;
- this pattern is similar at each speed.

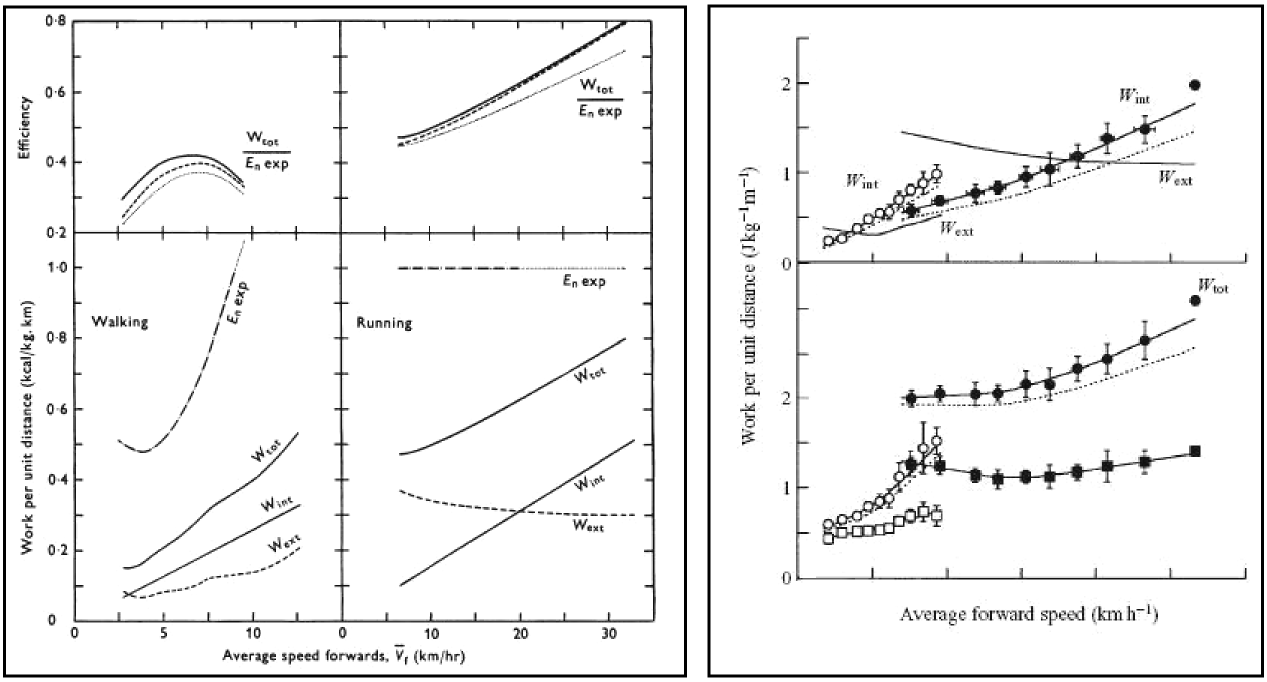


Figure 10.61. W_{ext}, W_{int} and W_{tot} as a function of speed, in Cavagna et al. (1983) (on the left) and in Willems et al. (1995) (on the right).

Specific results of the statistical analysis (with relevance) are shown in par. 7.3, Table 10.28 and 10.29 (see Appendix 10.1).

9.2.3. Mechanical total work as a function of gradient

For males and females aged 25 to 35, we analysed the mechanical total work in relation to gradient, too (Figure 10.62 and 10.63).

9.2.3.1. Mechanical total work in gradient walking

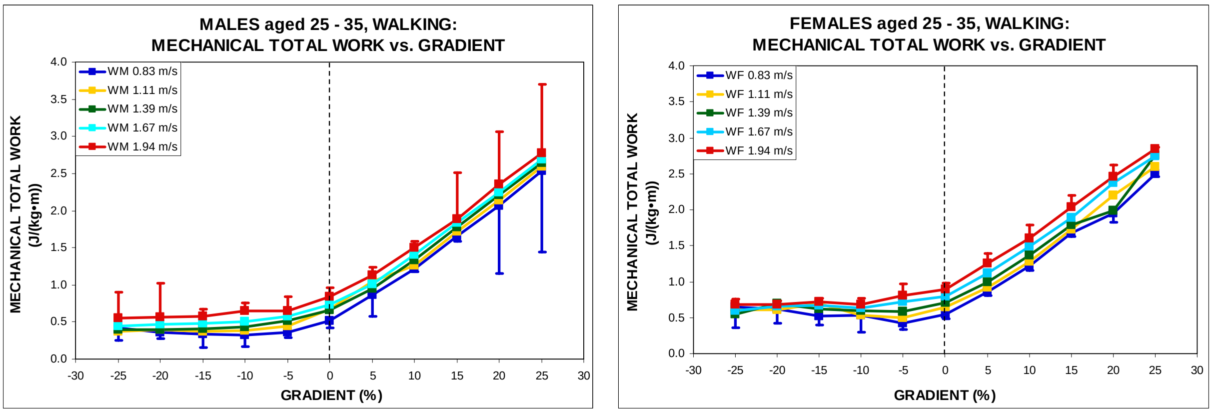


Figure 10.62. Mechanical total work as a function of gradient in walking,

males (on the left) and females (on the right).

- W_{tot} does not significantly change with gradient up to -5% (from 0.457 ± 0.185 to 0.512 ± 0.099 J/(kg·m), at all the investigated speeds, in males; and from 0.615 ± 0.145 to 0.605 ± 0.103 J/(kg·m), in females);
- above this gradient, W_{tot} is an upward linear function of gradient ($p<0.001$);
- this pattern is evident both in males (left graph) and females (right graph);
- as previously demonstrated, from -25 to -5%, W_{tot} is little higher in females;
- furthermore, in this case, W_{tot} is primarily dependent on the corresponding and similar pattern of W_{ext} (see par. 5.3.3 and 5.5.3 above). The increase in W_{ext} contributes therefore to a resulting increase in W_{tot}.

Specific results of the statistical analysis (with relevance) are shown in par. 7.5, Table 10.30 (see Appendix 10.1).

9.2.3.2. Mechanical total work in gradient running

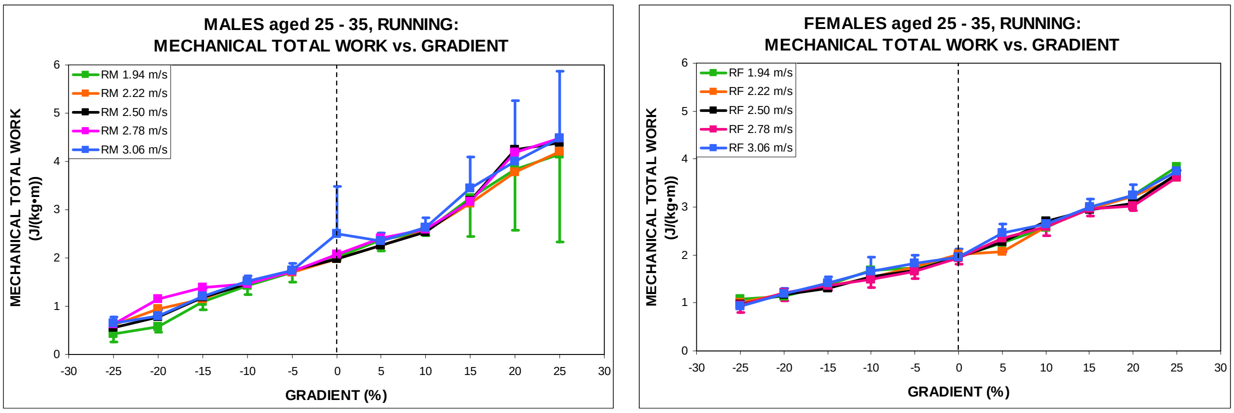


Figure 10.63. Mechanical total work as a function of gradient in running, males (on the left) and females (on the right).

- W_{tot} is an upward linear function of gradient ($p<0.001$);
- this pattern is evident both in males (left graph) and females (right graph) and it occurs at each speed;
- on average, from -25 to -15%, W_{tot} is higher in females (1.176 ± 0.134 J/(kg·m), at all the investigated speeds) than in males (0.994 ± 0.138 J/(kg·m));
- in males, it is important to underline the high standard deviations at the higher gradients (from 15 to 25%).

Specific results of the statistical analysis (with relevance) are shown in par. 7.6, Table 10.31 (see Appendix 10.1). Statistical analysis was not applied to female data, because of some discarded tests (see also chapters 5, par. 1.1.2, and 6, par. 2.2).

10. CONCLUSION

All our results concur with literature data (Cavagna et al., 1964; 1976; 1977; 1983; 1986; Thorstensson et al., 1987; Cavagna et al., 1991; Ardigò, 1992; Minetti et al., 1993; 1994; 1995; Willems et al., 1995; Cavagna et al., 1997; Minetti, 1998; Schepens et al., 1998; 2001; Saibene et al., 2003; Cavagna et al., 2008b). They also complete and extend this data because, in our study, we have taken account both lower and higher gradients (from -25 to -15% in downhill condition, and from 15 to 25% in uphill condition; Minetti et al., 1993).

As a result, each (simplex and complex) biomechanical variable permitted us to fully describe and investigate the mechanics of walking and running as gender, age, speed and gradient change.

Particularly, we can conclude that:

a) as far as **stride frequency** is concerned: both in walking and running, SF is similar in males and females; it is slightly dependent on age; it is highly dependent on speed ($p<0.001$); yet, it is independent of gradient;

b) as far as **stride length** is concerned: both in walking and running, SL is similar in males and females; it is slightly dependent on age; it is highly dependent on speed ($p<0.001$); yet, it is independent of gradient;

c) as far as **duty factor** is concerned: both in walking and running, DF is similar in males and females; it is independent of age; it is highly dependent on speed ($p<0.001$); furthermore, it is independent of gradient;

d) as far as **mechanical external work** is concerned: both in walking and running, W_{ext} is similar in males and females; it is slightly dependent on age. In walking (level and gradient condition), it is highly dependent on both speed and gradient ($p<0.001$). Yet, in running (level and gradient condition), it is independent of speed, but it is highly dependent on gradient ($p<0.001$);

e) as far as **energy recovery percentage** is concerned: in level walking, R is similar in males and females; it is slightly dependent on both age and speed; in gradient walking, it is not completely similar in males and females; finally, it is highly dependent on gradient;

f) as far as **mechanical internal work** is concerned: both in walking and running, W_{int} is similar in males and females; it is independent of age; it is highly dependent on speed ($p<0.001$); yet, it is independent of gradient;

g) as far as **mechanical total work** is concerned: both in walking and running, W_{tot} is similar in males and females; it is slightly dependent on age; it is highly dependent on both speed and gradient ($p<0.001$).

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Appendix 10.1

STATISTICAL ANALYSIS

1. INTRODUCTION

In this appendix, there are cross-section tables showing statistical results (with relevances) in each testing condition. They are arranged in the same order as already presented in chapter 10.

As previously presented, the alpha test level set for statistical significance was 0.05. If there is no significance (p=NS), the corresponding space in empty. In all the other cases, the significance is specified: in blue for males and in pink for females.

2. STRIDE FREQUENCY

2.1. Level walking: stride frequency as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show significant results in SF as a function of age at each speed.

LEVEL WALKING at 0.83 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**			$p<0.05$			
14 - 17 y		**Males** **Females**					
18 - 24 y			**Males** **Females**				
25 - 35 y	$p<0.05$			**Males** **Females**			
36 - 45 y					**Males** **Females**		
46 - 55 y						**Males** **Females**	
56 - 65 y							**Males** **Females**

Table 10.1a. SF as a function of age in level walking at 0.83 m/s (males and females).

Importantly, in level walking at 0.83 m/s, it must highlight the low (or nil) consistency of such reported significances, quite differently to what happens at the most other speeds.

LEVEL WALKING at 1.11 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**		$p<0.01$	$p<0.001$	$p<0.01$ $p<0.05$		
14 - 17 y		**Males** **Females**					
18 - 24 y	$p<0.01$		**Males** **Females**				
25 - 35 y	$p<0.001$			**Males** **Females**			
36 - 45 y	$p<0.01$ $p<0.05$				**Males** **Females**		
46 - 55 y						**Males** **Females**	
56 - 65 y							**Males** **Females**

Table 10.1b. SF as a function of age in level walking at 1.11 m/s (males and females).

LEVEL WALKING at 1.39 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**	$p<0.05$				$p<0.05$	
14 - 17 y	$p<0.05$	**Males** **Females**					
18 - 24 y			**Males** **Females**				
25 - 35 y				**Males** **Females**			
36 - 45 y					**Males** **Females**		
46 - 55 y	$p<0.05$					**Males** **Females**	
56 - 65 y							**Males** **Females**

Table 10.1c. SF as a function of age in level walking at 1.39 m/s (males and females).

Importantly, in level walking at 1.39 m/s, it must highlight the low (or nil) consistency of such reported significances, quite differently to what happens at the most other speeds.

LEVEL WALKING at 1.67 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Males** **Females**	$p<0.05$		$p<0.01$ $p<0.001$	$p<0.05$ $p<0.01$	$p<0.01$	
14 - 17 y	$p<0.05$	**Males** **Females**					
18 - 24 y			**Males** **Females**				
25 - 35 y	$p<0.01$ $p<0.001$			**Males** **Females**			
36 - 45 y	$p<0.05$ $p<0.01$				**Males** **Females**		
46 - 55 y	$p<0.01$					**Males** **Females**	
56 - 65 y							**Males** **Females**

Table 10.1d. SF as a function of age in level walking at 1.67 m/s (males and females).

LEVEL WALKING at 1.94 m/s	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 -45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males** **Females**	p<0.001 p<0.01		p<0.001 p<0.001	p<0.001 p<0.01	p<0.001	
14 - 17 y	p<0.001 p<0.01	**Males** **Females**					
18 - 24 y			**Males** **Females**				
25 - 35 y	p<0.001 p<0.001			**Males** **Females**			
36 - 45 y	p<0.001 p<0.01				**Males** **Females**		
46 - 55 y	p<0.001					**Males** **Females**	
56 - 65 y							**Males** **Females**

Table 10.1e. SF as a function of age in level walking at 1.94 m/s (males and females).

2.2. Level running: stride frequency as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, both in males and females, there is no significance in SF as a function of age.

SUBJECTS	LEVEL RUNNING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.2. SF as a function of age in level running (males and females).

2.3. Level walking and running: stride frequency as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that walking is highly different to running (p<0.001).

Moreover, both in males and females, there is a high significance in SF as a function of speed in each age group (p<0.001 in walking and p<0.05 in running). This pattern occurs at each speed. In females, there is an only exception between 2.50 and 2.78 m/s.

SUBJECTS	LEVEL WALKING in each age group	LEVEL RUNNING in each age group
Males	p<0.001 comparing all speeds	p<0.05 comparing all speeds
Females	p<0.001 comparing all speeds	p<0.05 comparing all speeds

Table 10.3. SF as a function of speed in level walking and running (males and females).

2.4. Walking: stride frequency as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males (excepted at 1.39 m/s) and females, there are significant results in SF as a function of gradient at each speed.

GRADIENT WALKING at 0.83 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**			$p<0.001$	$p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.001$ $p<0.05$	$p<0.001$
-20%		**Males** **Females**					$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$			
-15%			**Males** **Females**		$p<0.01$	$p<0.001$	$p<0.001$	$p<0.001$			
-10%	$p<0.001$			**Males** **Females**		$p<0.01$ $p<0.05$	$p<0.01$ $p<0.001$	$p<0.001$ $p<0.05$			
-5%	$p<0.001$		$p<0.01$		**Males** **Females**			$p<0.001$			
0%	$p<0.001$ $p<0.01$		$p<0.001$	$p<0.01$ $p<0.05$		**Males** **Females**					
5%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$	$p<0.01$ $p<0.001$			**Males** **Females**				
10%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$	$p<0.001$ $p<0.05$	$p<0.001$			**Males** **Females**			
15%	$p<0.001$								**Males** **Females**		
20%	$p<0.001$ $p<0.05$									**Males** **Females**	
25%	$p<0.001$										**Males** **Females**

Table 10.4a. SF as a function of gradient in walking at 0.83 m/s (males and females).

GRADIENT WALKING at 1.11 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**			$p<0.05$	$p<0.001$	$p<0.001$	$p<0.05$ $p<0.001$	$p<0.05$ $p<0.001$	$p<0.001$	$p<0.001$	
-20%		**Males** **Females**			$p<0.05$	$p<0.05$	$p<0.05$			$p<0.01$	
-15%			**Males** **Females**		$p<0.01$	$p<0.05$ $p<0.001$	$p<0.01$ $p<0.05$	$p<0.05$			
-10%	$p<0.05$			**Males** **Females**			$p<0.001$ $p<0.001$	$p<0.001$	$p<0.05$		
-5%	$p<0.001$	$p<0.05$	$p<0.01$		**Males** **Females**		$p<0.001$	$p<0.01$			
0%	$p<0.001$	$p<0.05$	$p<0.05$ $p<0.001$			**Males** **Females**					
5%	$p<0.05$ $p<0.001$	$p<0.05$	$p<0.01$ $p<0.05$	$p<0.001$ $p<0.001$	$p<0.001$		**Males** **Females**				
10%	$p<0.05$ $p<0.001$		$p<0.05$	$p<0.001$	$p<0.01$			**Males** **Females**			
15%	$p<0.001$			$p<0.05$					**Males** **Females**		
20%	$p<0.001$	$p<0.01$								**Males** **Females**	
25%											**Males** **Females**

Table 10.4b. SF as a function of gradient in walking at 1.11 m/s (males and females).

GRADIENT WALKING at 1.39 m/s	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	Females				$p<0.05$		$p<0.01$	$p<0.01$	$p<0.001$		
-20%		Females	$p<0.001$			$p<0.01$	$p<0.01$	$p<0.001$	$p<0.001$		
-15%		$p<0.001$	Females				$p<0.01$	$p<0.05$			
-10%				Females				$p<0.001$			
-5%	$p<0.05$				Females		$p<0.01$				
0%		$p<0.01$				Females	$p<0.001$				
5%	$p<0.01$	$p<0.01$	$p<0.01$		$p<0.01$	$p<0.001$	Females				
10%	$p<0.01$	$p<0.001$	$p<0.05$	$p<0.001$				Females			
15%	$p<0.001$	$p<0.001$							Females		
20%										Females	
25%											Females

Table 10.4c. SF as a function of gradient in walking at 1.39 m/s (females).

GRADIENT WALKING at 1.67 m/s	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	Males Females				$p<0.05$						
-20%		Males Females									
-15%			Males Females	$p<0.05$			$p<0.001$				
-10%			$p<0.05$	Males Females	$p<0.001$						
-5%	$p<0.05$			$p<0.001$	Males Females						
0%						Males Females					
5%			$p<0.001$				Males Females				
10%								Males Females			
15%									Males Females		
20%										Males Females	
25%											Males Females

Table 10.4d. SF as a function of gradient in walking at 1.67 m/s (males and females).

GRADIENT WALKING at 1.94 m/s	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	Males Females										
-20%		Males Females			$p<0.01$	$p<0.05$					
-15%			Males Females								
-10%				Males Females							
-5%		$p<0.01$			Males Females					$p<0.001$	
0%		$p<0.05$				Males Females				$p<0.001$	
5%							Males Females				
10%								Males Females	$p<0.001$		
15%								$p<0.001$	Males Females		
20%					$p<0.001$					Males Females	
25%					$p<0.001$						Males Females

Table 10.4e. SF as a function of gradient in walking at 1.94 m/s (males and females).

Importantly, in gradient walking at 1.67 and 1.94 m/s, it must highlight the low (or nil) consistency of such reported significances, quite differently to what happens at the other speeds.

2.5. Running: stride frequency as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show no significant results in SF as a function of gradient at each speed.

SUBJECTS	GRADIENT RUNNING at each speed
Males	p=NS comparing all gradients

Table 10.5. SF as a function of gradient in running (males).

3. DUTY FACTOR

3.1. Level walking: duty factor as a function of age

In males, one-way ANOVA and the post-*hoc* Bonferroni test show that there is no significant changes in DF as a function of age.

SUBJECTS	LEVEL WALKING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.6a. DF as a function of age in level walking (males and females).

However, in females, there are significant results in DF as a function of age at the higher speed (1.94 m/s, p<0.001).

LEVEL WALKING at 1.94 m/s	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Females**	p<0.001	p<0.001	p<0.001	p<0.001	p<0.001	p<0.001
14 - 17 y	p<0.001	**Females**					
18 - 24 y	p<0.001		**Females**				
25 - 35 y	p<0.001			**Females**			
36 - 45 y	p<0.001				**Females**		
46 - 55 y	p<0.001					**Females**	
56 - 65 y	p<0.001						**Females**

Table 10.6b. DF as a function of age in level walking at 1.94 m/s (females).

3.2. Level running: duty factor as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that there is no significance in DF as a function of age in level running. This pattern occurs both in males and females.

SUBJECTS	LEVEL RUNNING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.7. DF as a function of age in level running (males and females).

3.3. Level walking and running: duty factor as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that walking is highly different to running (p<0.001).

Moreover, there is a high significance in DF regarding speed (p<0.001). This pattern occurs both in males and females.

SUBJECTS	LEVEL WALKING AND RUNNING in each age group
Males	p<0.001 comparing all speeds
Females	p<0.001 comparing all speeds

Table 10.8. DF as a function of speed in level walking and running (males and females).

3.4. Walking: duty factor as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show significant results in DF as a function of gradient at each speed. This pattern occurs both in males (excepted at the lower speeds) and females.

GRADIENT WALKING at 0.83 m/s	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	**Females**				p<0.01						
-20%		**Females**									
-15%			**Females**								
-10%				**Females**							
-5%	p<0.01				**Females**						
0%						**Females**					
5%							**Females**				
10%								**Females**			
15%									**Females**		
20%										**Females**	
25%											**Females**

Table 10.9a. DF as a function of gradient in walking at 0.83 m/s (females).

Importantly, in gradient walking at 0.83 m/s, it must highlight the low (or nil) consistency of such reported significances, quite differently to what happens at the most other speeds.

GRADIENT WALKING at 1.11m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Females**					$p<0.01$	$p<0.05$	$p<0.01$	$p<0.01$	$p<0.01$	$p<0.05$
-20%		**Females**				$p<0.01$	$p<0.01$		$p<0.01$		
-15%			**Females**								
-10%				**Females**							
-5%					**Females**						
0%	$p<0.01$	$p<0.01$				**Females**					
5%	$p<0.05$	$p<0.01$					**Females**				
10%	$p<0.01$							**Females**			
15%	$p<0.01$	$p<0.01$							**Females**		
20%	$p<0.01$									**Females**	
25%	$p<0.05$										**Females**

Table 10.9b. DF as a function of gradient in walking at 1.11 m/s (females).

GRADIENT WALKING at 1.39 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**					$p<0.05$	$p<0.05$		$p<0.01$		
-20%		**Males** **Females**	$p<0.01$								
-15%		$p<0.01$	**Males** **Females**			$p<0.001$	$p<0.001$				
-10%				**Males** **Females**					$p<0.001$		
-5%					**Males** **Females**				$p<0.05$		
0%	$p<0.05$		$p<0.001$			**Males** **Females**					
5%	$p<0.05$		$p<0.001$				**Males** **Females**				
10%								**Males** **Females**			
15%	$p<0.01$			$p<0.001$	$p<0.05$				**Males** **Females**		
20%										**Males** **Females**	
25%											**Males** **Females**

Table 10.9c. DF as a function of gradient in walking at 1.39 m/s (males and females).

GRADIENT WALKING at 1.67 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**				$p<0.01$	$p<0.05$	$p<0.01$		$p<0.01$	$p<0.001$	$p<0.001$
-20%		**Males** **Females**				$p<0.01$	$p<0.05$	$p<0.05$		$p<0.001$	
-15%			**Males** **Females**								
-10%				**Males** **Females**					$p<0.05$		
-5%	$p<0.01$				**Males** **Females**			$p<0.05$			
0%	$p<0.05$	$p<0.01$				**Males** **Females**					
5%	$p<0.01$	$p<0.05$					**Males** **Females**				
10%		$p<0.05$			$p<0.05$			**Males** **Females**			
15%	$p<0.01$			$p<0.05$					**Males** **Females**		
20%	$p<0.001$	$p<0.001$								**Males** **Females**	
25%	$p<0.001$										**Males** **Females**

Table 10.9d. DF as a function of gradient in walking at 1.67 m/s (males and females).

GRADIENT WALKING at 1.94 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**		$p<0.01$		$p<0.001$	$p<0.001$ $p<0.05$	$p<0.01$	$p<0.01$	$p<0.001$	$p<0.001$	$p<0.001$
-20%		**Males** **Females**				$p<0.01$	$p<0.01$ $p<0.05$	$p<0.001$ $p<0.01$			
-15%	$p<0.01$		**Males** **Females**								
-10%				**Males** **Females**							
-5%	$p<0.001$				**Males** **Females**		$p<0.001$	$p<0.01$			
0%	$p<0.001$ $p<0.05$	$p<0.01$				**Males** **Females**					
5%	$p<0.01$	$p<0.01$ $p<0.05$			$p<0.001$		**Males** **Females**				
10%	$p<0.01$	$p<0.001$ $p<0.01$			$p<0.01$			**Males** **Females**			
15%	$p<0.001$								**Males** **Females**		
20%	$p<0.001$									**Males** **Females**	
25%	$p<0.001$										**Males** **Females**

Table 10.9e. DF as a function of gradient in walking at 1.94 m/s (males and females).

3.5. Running: duty factor as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show significant results in DF as a function of gradient only at the speeds of 2.22 m/s ($p<0.05$) and 3.06 m/s ($p<0.05$ and $p<0.01$). However, there is no significance at the other speeds.

GRADIENT RUNNING at 2.22 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males**										
-20%		**Males**									
-15%			**Males**								
-10%				**Males**				$p<0.05$			
-5%					**Males**						
0%						**Males**					
5%							**Males**				
10%				$p<0.05$				**Males**			
15%									**Males**		
20%										**Males**	
25%											**Males**

Table 10.10a. DF as a function of gradient in running at 2.22 m/s (males).

GRADIENT RUNNING at 3.06 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males**										
-20%		**Males**									
-15%			**Males**				$p<0.05$		$p<0.05$		
-10%				**Males**					$p<0.01$		
-5%					**Males**						
0%						**Males**					
5%			$p<0.05$				**Males**				
10%								**Males**			
15%			$p<0.05$	$p<0.01$					**Males**		
20%										**Males**	
25%											**Males**

Table 10.10b. DF as a function of gradient in running at 3.06 m/s (males).

Importantly, it must highlight the low consistency of such reported significances.

4. MECHANICAL EXTERNAL WORK

4.1. Level walking: mechanical external work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that there is no significance between W_{ext} and age in males.

SUBJECTS	**LEVEL WALKING at each speed**
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.11a. W_{ext} as a function of age in level walking (males and females).

However, in females, there is a relationship at the higher speed of 1.94 m/s ($p<0.001$).

LEVEL WALKING at 1.94 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Females**	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$
14 - 17 y	$p<0.001$	**Females**					
18 - 24 y	$p<0.001$		**Females**				
25 - 35 y	$p<0.001$			**Females**			
36 - 45 y	$p<0.001$				**Females**		
46 - 55 y	$p<0.001$					**Females**	
56 - 65 y	$p<0.001$						**Females**

Table 10.11b. W_{ext} as a function of age in level walking at 1.94 m/s (females).

4.2. Level walking: mechanical external work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that walking is highly different to running ($p<0.001$).

Moreover, there are significant results in W_{ext} as a function of speed ($p<0.001$). This pattern occurs both in males and females.

SUBJECTS	LEVEL WALKING in each age group
Males	$p<0.001$ comparing all speeds
Females	$p<0.001$ comparing all speeds

Table 10.12. W_{ext} as a function of speed in level walking (males and females).

There are only few exceptions:

a) in males, p is not significant between 1.11 and 1.39 m/s;

b) in males, p is < 0.01 between 0.83 and 1.11 m/s;

c) in females, p is < 0.01 between 1.11 and 1.39 m/s.

4.3. Walking: mechanical external work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that there is a high significance in W_{ext} as a function of gradient ($p<0.001$). This pattern is similar both in males and females, at each speed.

GRADIENT WALKING at each speed	-25%	-20%	-15%	-10%	-5%	0%	5%	10%	15%	20%	25%
-25%	**Males** **Females**					$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-20%		**Males** **Females**				$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-15%			**Males** **Females**			$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-10%				**Males** **Females**		$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-5%					**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
0%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
5%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
10%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
15%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
20%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$
25%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**

Table 10.13. W_{ext} as a function of gradient in walking (males and females).

4.4. Level running: mechanical external work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that there is only very little significance between W_{ext} and age in males (excepted for $p<0.001$ at the higher speed).

SUBJECTS	LEVEL RUNNING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.14a. W_{ext} as a function of age in level running (males and females).

LEVEL RUNNING at 3.06 m/s	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males**						
14 - 17 y		**Males**					
18 - 24 y			**Males**				
25 - 35 y				**Males**			p<0.001
36 - 45 y					**Males**		
46 - 55 y						**Males**	
56 - 65 y				p<0.001			**Males**

Table 10.14b. W_{ext} as a function of age in level running at 3.06 m/s (males).

Importantly, it must highlight the low (or nil) consistency of such reported significances. However, in females, there is no significant change in W_{ext} as a function of speed.

4.5. Level running: mechanical external work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, in males and females, there is no significance in W_{ext} regarding speed.

SUBJECTS	LEVEL RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 10.15. W_{ext} as a function of speed in level running (males and females).

4.6. Running: mechanical external work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, in males and females, there is a high significance in W_{ext} as a function of gradient ($p<0.001$). This pattern is similar at each speed.

SUBJECTS	GRADIENT RUNNING at each speed
Males	p<0.001 comparing all gradients
Females	p<0.001 comparing all gradients

Table 10.16. W_{ext} as a function of gradient in running (males and females).

5. ENERGY RECOVERY PERCENTAGE

5.1. Level walking: energy recovery percentage as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, in males, there is no significance in R as a function of age at all speeds.

SUBJECTS	LEVEL WALKING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.17a. R as a function of age in level walking (males and females).

However, in females, there is a little significance at the speeds of 1.39 and 1.94 m/s ($p<0.05$ and $p<0.001$, respectively).

LEVEL WALKING at 1.39 m/s	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Females**						$p<0.05$
14 - 17 y		**Females**					
18 - 24 y			**Females**				
25 - 35 y				**Females**			
36 - 45 y					**Females**		
46 - 55 y						**Females**	
56 - 65 y	$p<0.05$						**Females**

Table 10.17b. R as a function of age in level walking at 1.39 m/s (females).

Importantly, in gradient walking at 1.39 m/s, it must highlight the low (or nil) consistency of such reported significances, quite differently to what happens at the other speed.

LEVEL WALKING at 1.94 m/s	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Females**	$p<0.05$		$p<0.001$	$p<0.05$	$p<0.01$	
14 - 17 y	$p<0.05$	**Females**					
18 - 24 y			**Females**				
25 - 35 y	$p<0.001$			**Females**			
36 - 45 y	$p<0.05$				**Females**		
46 - 55 y	$p<0.01$					**Females**	
56 - 65 y							**Females**

Table 10.17c. R as a function of age in level walking at 1.94 m/s (females).

5.2. Level walking: energy recovery percentage as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, in each male age groups, in some cases, there is a high significance in R as a function of speed ($p<0.001$).

LEVEL WALKING	**0.83 m/s**	**1.11 m/s**	**1.39 m/s**	**1.67 m/s**	**1.94 m/s**
0.83 m/s	**Males**	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$
1.11 m/s	$p<0.001$	**Males**			
1.39 m/s	$p<0.001$		**Males**		
1.67 m/s	$p<0.001$			**Males**	
1.94 m/s	$p<0.001$				**Males**

Table 10.18a. R as a function of speed in level walking (males).

However, in females, there is no significance between R and speed in each age group.

SUBJECTS	**LEVEL WALKING in each age group**
Females	p=NS comparing all speeds

Table 10.18b. R as a function of speed in level walking (females).

5.3. Walking: energy recovery percentage as a function of gradient

One-way ANOVA for related measures and the post-*hoc* *t*-test (with Bonferroni correction) show that, in males and females, there is a high significance in R as a function of gradient.

GRADIENT WALKING at each speed	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**				$p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-20%		**Males** **Females**			$p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-15%			**Males** **Females**		$p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-10%				**Males** **Females**	$p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-5%	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
0%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.001$	$p<0.001$ $p<0.001$
5%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.05$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
10%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
15%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
20%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$
25%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**

Table 10.19. R as a function of gradient in walking (males and females).

6. MECHANICAL INTERNAL WORK

6.1. Level walking: mechanical internal work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, both in males and females, there is no significance in W_{int} as a function of age.

SUBJECTS	LEVEL WALKING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.20. W_{int} as a function of age in level walking (males and females).

6.2. Level running: mechanical internal work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, both in males and females, there is no significance in W_{int} as a function of age.

SUBJECTS	LEVEL RUNNING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.21. W_{int} as a function of age in level running (males and females).

6.3. Level walking: mechanical internal work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired t-test (with Bonferroni correction) show that, both in males and females, there is a high significance in W_{int} regarding speed ($p<.0.001$).

SUBJECTS	LEVEL WALKING in each age group
Males	p<0.001 comparing all speeds
Females	p<0.001 comparing all speeds

Table 10.22. W_{int} as a function of speed in level walking (males and females).

There are only few exceptions:

a) in males, p not significant between 1.11 and 1.39 m/s;

b) in females, p is < 0.01 between 1.11 and 1.39 m/s;

c) in females, p is < 0.01 between 1.11 and 1.39 m/s.

6.4. Level running: mechanical internal work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is a high significance in W_{int} regarding speed ($p<.0.001$).

SUBJECTS	LEVEL RUNNING in each age group
Males	$p<0.001$ comparing all speeds
Females	$p<0.001$ comparing all speeds

Table 10.23. W_{int} as a function of speed in level running (males and females).

There are only few exceptions:

a) in males, p is not significant between 1.94 and 2.22 m/s and between 2.78 and 3.06 m/s;

b) in males, p is < 0.01 between 2.22 and 2.50 m/s;

c) in females, p is not significant between 1.94 and 2.22 m/s;

d) in females, p is < 0.01 between 1.94 and 2.50 m/s.

6.5. Walking: mechanical internal work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is no significance in W_{int} regarding gradient.

SUBJECTS	GRADIENT WALKING at each speed
Males	p=NS comparing all gradients
Females	p=NS comparing all gradients

Table 10.24. W_{int} as a function of gradient in walking (males and females).

6.6. Running: mechanical internal work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is no significance in W_{int} as a function of gradient at each speed.

SUBJECTS	GRADIENT RUNNING at each speed
Males	p=NS comparing all gradients
Females	p=NS comparing all gradients

Table 10.25a. W_{int} as a function of gradient in running (males and females).

There is only an exception: in males, p is < 0.05 at the higher speed of 3.06 m/s.

GRADIENT RUNNING at 3.06 m/s	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males**								$p<0.05$	$p<0.05$	$p<0.05$
-20%		**Males**							$p<0.05$	$p<0.05$	$p<0.05$
-15%			**Males**						$p<0.05$	$p<0.05$	$p<0.05$
-10%				**Males**					$p<0.05$	$p<0.05$	$p<0.05$
-5%					**Males**			$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$
0%						**Males**		$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$
5%							**Males**	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.01$
10%					$p<0.05$	$p<0.05$	$p<0.05$	**Males**	$p<0.05$	$p<0.05$	$p<0.05$
15%	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	**Males**	$p<0.05$	$p<0.05$
20%	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	**Males**	$p<0.05$
25%	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.05$	$p<0.01$	$p<0.05$	$p<0.05$	$p<0.05$	**Males**

Table 10.25b. W_{int} as a function of gradient in running at 3.06 m/s (males).

However, it is important to highlight the low consistency of such reported significances.

7. MECHANICAL TOTAL WORK

7.1. Level walking: mechanical total work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, in males, there is no significance in W_{tot} regarding age at each speed.

SUBJECTS	**LEVEL WALKING at each speed**
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.26a. W_{tot} as a function of age in level walking (males and females).

However, in females, there is a statistical significance only at the higher speed of 1.94 m/s.

LEVEL WALKING at 1.94 m/s	**6 - 13 y**	**14 - 17 y**	**18 - 24 y**	**25 - 35 y**	**36 - 45 y**	**46 - 55 y**	**56 - 65 y**
6 - 13 y	**Females**	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$	$p<0.001$
14 - 17 y	$p<0.001$	**Females**					
18 - 24 y	$p<0.001$		**Females**				
25 - 35 y	$p<0.001$			**Females**			
36 - 45 y	$p<0.001$				**Females**		
46 - 55 y	$p<0.001$					**Females**	
56 - 65 y	$p<0.001$						**Females**

Table 10.26b. W_{tot} as a function of age in level walking at 1.94 m/s (females).

7.2. Level running: mechanical total work as a function of age

One-way ANOVA and the post-*hoc* Bonferroni test show that, both in males and females, there is no significance in W_{tot} regarding age at each speed.

SUBJECTS	LEVEL RUNNING at each speed
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 10.27. W_{tot} as a function of age in level running (males and females).

7.3. Level walking: mechanical total work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is a high significance in W_{tot} regarding speed ($p<0.001$).

SUBJECTS	LEVEL WALKING in each age group
Males	p<0.001 comparing all speeds
Females	p<0.001 comparing all speeds

Table 10.28. W_{tot} as a function of speed in level walking (males and females).

7.4. Level running: mechanical total work as a function of speed

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is no significance in W_{tot} regarding speed. There is only an exception: in females, p is < 0.01 between 2.78 and 3.06 m/s.

SUBJECTS	LEVEL RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 10.29. W_{tot} as a function of speed in level running (males and females).

7.5. Walking: mechanical total work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that there is a high significance in W_{tot} as a function of gradient ($p<0.001$). This pattern is similar both in males and females, at each speed.

GRADIENT WALKING at each speed	**-25%**	**-20%**	**-15%**	**-10%**	**-5%**	**0%**	**5%**	**10%**	**15%**	**20%**	**25%**
-25%	**Males** **Females**					$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-20%		**Males** **Females**				$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-15%			**Males** **Females**			$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-10%				**Males** **Females**		$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
-5%					**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
0%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
5%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
10%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
15%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$
20%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**	$p<0.001$ $p<0.001$
25%	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	$p<0.001$ $p<0.001$	**Males** **Females**

Table 10.30. W_{tot} as a function of gradient in walking (males and females).

7.6. Running: mechanical total work as a function of gradient

One-way ANOVA for related measures and the post-*hoc* paired *t*-test (with Bonferroni correction) show that, both in males and females, there is a high significance in W_{tot} as a function of gradient ($p<0.001$). This pattern is similar at each speed.

SUBJECTS	**GRADIENT RUNNING at each speed**
Males	$p<0.001$ comparing all gradiens
Females	$p<0.001$ comparing all gradients

Table 10.31. W_{tot} as a function of gradient in running (males and females).

Chapter 11

FOURIER ANALYSIS TO CALCULATE BIOMECHANICAL VARIABLES

1. BIOMECHANICAL VARIABLES ANALIZED BY FOURIER ANALYSIS

1.1. Introduction

The mathematical innovative method (based upon Fourier analysis; see also chapter 6, par. 4 and 5) allowed us to measure some complex biomechanical variables.

Particularly, we used harmonic coefficients and phases to evaluate both the mechanical external work (W_{ext}) and the energy recovery percentage (R). Other detailed information about these biomechanical variables have been already discussed in chapter 10.

1.2. Single steps in this mathematical analysis

In this paragraph, all steps we followed to appreciate both mechanical external work and energy recovery percentage are presented and discussed.

A. Because of the periodicity of the 3D displacements of the body centre of mass, the basic movement time was defined in a circumference: from 0 to 360 degrees (t expressed in degree), and/or from 0 to 2π radiant (t expressed in radiant).

Moreover, the derived movement time *t'* (in sec) was calculated as:

$$t' = \frac{t}{(2\pi) \bullet Fr} \quad \text{[Eq. 11.1]}$$

where *t* represents the basic movement time (in radiant); and *Fr* the sampling acquisition frequency (in Hz; see also chapter 4, par. 3.4.4). This value was assumed to keep constant in each testing condition.

B. Symmetrical harmonic coefficients (A) and phases (φ) were used to measure the movement/trajectory of the BCOM along each movement direction: x-axis (anterior/posterior), y-axis (vertical) and z-axis (medial/lateral). These displacements were calculated according to rules and principles both from trigonometric mathematics and Fourier analysis and are expressed in mm:

$$x = Ax2 \bullet \sin(2t + \varphi 2) + Ax4 \bullet \sin(4t + \varphi 4) + Ax6 \bullet \sin(6t + \varphi 6) \quad \text{[Eq. 11.2a]}$$

$$y = Ay2 \bullet \sin(2t + \varphi 2) + Ay4 \bullet \sin(4t + \varphi 4) + Ay6 \bullet \sin(6t + \varphi 6) \quad \text{[Eq. 11.2b]}$$

$$z = Az1 \cdot \sin(t + \varphi 1) + Az3 \cdot \sin(3t + \varphi 3) + Az5 \cdot \sin(5t + \varphi 5) \quad \text{[Eq. 11.2c]}$$

where *A* represents the amplitude coefficient (in mm); *t* the derived movement time (in radiant); and *φ* the phase coefficient (in radiant).

C. The vertical displacement (y) was used to calculate the gravitational potential energy (PE, expressed in J):

$$PE = y \cdot g \cdot m \quad \text{[Eq. 11.3]}$$

where *y* is the vertical displacement (converted from mm to m); *g* the gravity acceleration (= 9.81 m/sec^2); and *m* the body mass of each age group (in kg).

For more details about the characteristics of this type of energy, see also chapter 1, par. 4.2.

D. Furthermore, displacements along each movement direction (x, y and z) were used to obtain corresponding speeds (x-speed or v_x, y-speed or v_y, and z-speed or v_z, all expressed in m/s):

$$v_x = \frac{x_n - x_{n-1}}{dt} \quad \text{[Eq. 11.4a]}$$

$$v_y = \frac{y_n - y_{n-1}}{dt} \quad \text{[Eq. 11.4b]}$$

$$v_z = \frac{z_n - z_{n-1}}{dt} \quad \text{[Eq. 11.4c]}$$

where x_{n-1}, y_{n-1} and z_{n-1} represent the final trajectories of the BCOM, whereas x_n, y_n and z_n the initial trajectories; *dt* is the reciprocal number of the sampling frequency (in sec) obtained as:

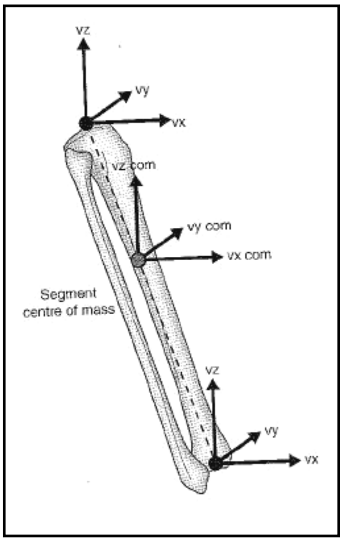
$$dt = \frac{1}{\frac{Fr}{90}} \quad \text{[Eq. 11.4d]}$$

E. Each component of the kinetic energy (KE_x, KE_y and KE_z, all expressed in J; Figure 11.1. Translational kinetic energy, in Richards (2008); Aleshinsky, 1986; Winter, 2005) was then calculated using all the previous information:

$$KE_x = \frac{1}{2} \cdot m \cdot (v_x + v)^2 \quad \text{[Eq. 11.5a]}$$

$$KE_y = \frac{1}{2} \cdot m \cdot (v_y)^2 \quad \text{[Eq. 11.5b]}$$

$$KE_Z = \frac{1}{2} \bullet m \bullet (v_Z)^2 \text{ [Eq. 11.5c]}$$



where *m* is the body mass of each group (in kg); v_x the speed in the forward movement direction (at each time, in m/sec); *v* the average speed progression; v_y the speed in the vertical movement direction (in m/sec); and v_z the speed in the medial/lateral movement direction (in m/sec).

For more details about the characteristics of this type of energy, see also chapter 1, par. 4.2.

F. Particularly, the total kinetic energy (Total KE, expressed in J) was calculated as the sum of KE_x and KE_z:

$$\text{TotalKE} = KE_X + KE_Z - \min KE_X \text{ [Eq. 11.6]}$$

where $minKE_x$ is the minimum forward kinetic energy value, in the circumference dimensions.

G. Moreover, the total potential energy (Total PE, expressed in J) was calculated as the sum of PE and KE_y:

$$\text{TotalPE} = PE + KE_Y - \min PE \text{ [Eq. 11.7]}$$

where *minPE* is the minimum potential energy value, in the circumference dimensions.

H. Consequently, the total energy (TE, expressed in J) was calculated as the sum of KE (Equation [11.6]) and PE (Equation [11.7]), which were clearly considered as two separate entities (see also chapter 1, par. 4.2):

$$TE = \text{TotalKE} + \text{TotalPE} \text{ [Eq. 11.8]}$$

I. Moreover, the mechanical external work (W_{ext}, expressed in J) sum was calculated as:

$$W_{ext}sum_{n-1} = \text{if}((TE_{n-1} - TE_n) > 0; \text{then}(TE_{n-1} - TE_n) + W_{ext}sum_n; W_{ext}sum_n) \text{ [Eq. 11.9]}$$

where $W_{ext}sum_{n-1}$ constitutes the mechanical external work at the last time, whereas the $W_{ext}sum_n$ the mechanical external work at the initial time. Clearly, TE represents the total energy.

J. As a result, the single kinetic energies derived from Fourier analysis (from Lissajous or L; KE_x from L; KE_y from L; KE_z from L, all expressed in J; Aleshinsky, 1986) were calculated as:

$$\mathrm{KE_x fromL} = \frac{1}{2} \bullet \mathrm{m} \bullet (\mathrm{v} + 2\pi \mathrm{Fr} \bullet (2\mathrm{Ax}2 \bullet \cos(2\mathrm{t} + \varphi 2)) + (4\mathrm{Ax}4 \bullet \cos(4\mathrm{t} + \varphi 4)) + (6\mathrm{Ax}6 \bullet \cos(6\mathrm{t} + \varphi 6)))^2$$

[Eq. 11.10a]

$$\mathrm{KE_y fromL} = \frac{1}{2} \bullet \mathrm{m} \bullet (\mathrm{v} + 2\pi \mathrm{Fr} \bullet (2\mathrm{Ay}2 \bullet \cos(2\mathrm{t} + \varphi 2)) + (4\mathrm{Ay}4 \bullet \cos(4\mathrm{t} + \varphi 4)) + (6\mathrm{Ay}6 \bullet \cos(6\mathrm{t} + \varphi 6)))^2$$

[Eq. 11.10b]

$$\mathrm{KE_z fromL} = \frac{1}{2} \bullet \mathrm{m} \bullet (\mathrm{v} + 2\pi \mathrm{Fr} \bullet (\mathrm{Az}1 \bullet \cos(\mathrm{t} + \varphi 1)) + (3\mathrm{Az}3 \bullet \cos(3\mathrm{t} + \varphi 3)) + (5\mathrm{Az}5 \bullet \cos(5\mathrm{t} + \varphi 5)))^2$$

[Eq. 11.10c]

K. The potential energy from Lissajous/Fourier analysis was calculated as:

$$\mathrm{PEfromL} = \mathrm{m} \bullet \mathrm{g} \bullet ((2\mathrm{Ay}2 \bullet \sin(2\mathrm{t} + \varphi 2)) + (4\mathrm{Ay}4 \bullet \cos(4\mathrm{t} + \varphi 4)) + (6\mathrm{Ay}6 \bullet \cos(6\mathrm{t} + \varphi 6)))^2$$

[Eq. 11.11]

L. Particularly, the total kinetic energy (Total KE) from Lissajous/Fourier was calculated as the sum of KE_x and KE_z:

$$\mathrm{TotalKEfromL} = \mathrm{KE_x fromL} + \mathrm{KE_z fromL} - \min \mathrm{KE_x fromL} \quad \text{[Eq. 11.12]}$$

where *minKE*$_x$*fromL* is the minimum forward kinetic energy value derived from Lissajous/Fourier mathematical analysis.

M. Moreover, the total potential energy (Total PE; Aleshinsky, 1986) from Lissajous/Fourier was calculated as the sum of PE and KE_y:

$$\mathrm{TotalPEfromL} = \mathrm{PEfromL} + \mathrm{KE_y fromL} - \min \mathrm{PEfromL} \quad \text{[Eq. 11.13]}$$

where *minPEfromL* is the minimum potential energy value.

N. Consequently, the total energy (Total TE) from Lissajous/Fourier was calculated as the sum of KE (Equation [11.12]) and PE (Equation [11.13]):

$$\mathrm{TotalTEfromL} = \mathrm{KEfromL} + \mathrm{PEfromL} \quad \text{[Eq. 11.14]}$$

O. The mechanical external work (W_{ext}) from Lissajous/Fourier sum was then calculated as:

$$\mathrm{W_{ext} sum_{n-1} fromL} = \mathrm{if}((\mathrm{TE_{n-1} fromL} - \mathrm{TE_n fromL}) > 0; \mathrm{then}(\mathrm{TE_{n-1}} - \mathrm{TE_n});0)$$

[Eq. 11.15a]

Furthermore, the forward mechanical external work (W_{ext}forward) from Lissajous/Fourier was calculated as:

$$W_{ext}\mathrm{forwardsum}_{n\text{-}1}\mathrm{fromL} = \mathrm{if}((KE_{n\text{-}1}\mathrm{fromL} - KE_{n}\mathrm{fromL}) > 0; \mathrm{then}(KE_{n\text{-}1} - KE_{n});0)$$

[Eq. 11.15b]

Finally, the vertical mechanical external work (W_{ext}vertical) from Lissajous/Fourier was calculated as:

$$W_{ext}\mathrm{verticalsum}_{n\text{-}1}\mathrm{fromL} = \mathrm{if}((PE_{n\text{-}1}\mathrm{fromL} - PE_{n}\mathrm{fromL}) > 0; \mathrm{then}(PE_{n\text{-}1} - PE_{n});0)$$

[Eq. 11.15c]

where $W_{ext}forwardsum_{n-1}$ and $W_{ext}verticalsum_{n-1}$ represent the mechanical external work (forward and vertical, respectively) at the last time.

All the previous equations ([11.15a], [11.15b] and [11.15c]) were then used to appreciate the energy recovery percentage (R), as a function of time:

$$R = \mathrm{if}((W_{ext}\mathrm{forwardsum} + W_{ext}\mathrm{verticalsum}) = 0; \frac{1 - W_{ext}\mathrm{sum}}{W_{ext}\mathrm{forwardsum} + W_{ext}\mathrm{verticalsum}})$$

[Eq. 11.16a]

$$\mathrm{Energy\,Recovery} = \frac{1 - \mathrm{abs.}(TE_{n\text{-}1} - TE_{n})}{\mathrm{abs.}(KE_{n\text{-}1} - KE_{n}) + \mathrm{abs.}(PE_{n\text{-}1} - PE_{n})}$$

[Eq. 11.16b]

where *TE*, *TotalKE* and *TotalPE* refer to the energy values derived from Lissajous/Fourier analysis.

The values of external work and energy recovery herein obtained took into account only the symmetrical coefficients pattern. However, it has been widely described that Fourier Series are generally characterised by both symmetrical and asymmetrical coefficients (see also chapter 6).

So, *what is the role of asymmetrical harmonic coefficients and phases?*

P. All the information described in points A - O were used to calculate the same measurements for the **asymmetrical coefficients**, as well.

Q. Of utmost importance, the displacements along each movement direction (both for symmetrical and asymmetrical coefficients) were then used to calculate the corresponding contributes of energy: total kinetic energy, total potential energy and total energy. Finally, these values were used to derive the mechanical external work and the energy recovery percentage (as a function of time).

R. In conclusion, using symmetrical and asymmetrical harmonic coefficients and phases derived from mathematical Lissajous/Fourier analysis, it becomes possible to definitively calculate:

- the mechanical external work (W_{ext}):

$$W_{ext} \frac{= \text{sumofW}_{ext}\text{sum}}{\frac{\text{m}}{\frac{\text{Fr}}{\text{v}}}} \quad \text{[Eq. 11.17]}$$

- the energy recovery percentage (R):

$$R = \frac{1 - \text{sumofW}_{ext}\text{sum}}{\text{sumofW}_{ext}\text{forward} + \text{sumofW}_{ext}\text{vertical}} \quad \text{[Eq. 11.18]}$$

The definitive values of external work and energy recovery percentage herein obtained took into account both symmetrical and asymmetrical coefficients pattern.

The general template we have used is contained in the enclosed CD (First Study, Chapter 11, Energies from Lissajous). Therefore, the parameters we have to insert in the Excel file in order to obtain all these information were:

a) symmetrical and asymmetrical harmonics coefficients;
b) stride frequency (in Hz) derived from file *.res (see also chapter 6, par. 2.1);
c) body mass of the subject or the age group subjects (in Kg);
d) speed of progression (in m/s) derived from the *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 (see also chapter 6, par. 2.1).

In this way, firstly it became possible to graph the pattern of each type of energy as a function of time (see Figure 11.2 and 11.3 below). Moreover, we calculated both mechanical external work (W_{ext}) and energy recovery percentage (R) using these values of energies. Particularly, each of these biomechanical variables has been calculated only in level conditions (both in males and females), both for walking and running, for each age group (see also chapter 5, par. 1.2). Thus, average values of body mass, speed and stride frequency have been considered.

In the sections below, we will focus on these biomechanical variables and their relationship with gender, age and speed. At the end, we will investigate the relationship between the cycle by cycle method (or discrete method) and the energy method (or mathematical/continuous method).

1.3. Potential and kinetic energy in walking

As previously presented in chapter 1 (par. 4.3.2; Figure 1.10 and 1.12), the **inverted pendulum model** accurately predicts the general pattern of mechanical energy fluctuations of the body for walking (Cavagna et al., 1963; 1976; Cavagna, 1977a; Cavagna et al., 1977b; Heglund et al., 1982; Alexander, 1984; di Prampero, 1985; Cavagna et al., 1988; Saibene, 1990; Mc Kinnon et al., 1993; Minetti et al., 1993; Donelan et al., 1997; Lejeune et al., 1998; Griffin et al., 1999; Cavagna et al., 2000; Preedy et al., 2001; Cavagna et al., 2002; Donelan et al., 2002b; Kuo, 2002; Gage et al., 2004; Hallemans et al., 2004; Alexander, 2005; Cavagna et al., 2005; Ortega et al., 2005; Robilliard et al., 2005; Biewener, 2006; Hoyt et al., 2006; Kuo, 2007; Segers et al., 2007; Sawicki et al., 2008; Usherwood et al., 2008; Winiarski, 2008; Houdijk et al., 2009; Whittington et al., 2009). However, this conceptual model should not be taken literally (Zatsiorsky, 2002; Geyer et al., 2005; Kuo et al., 2005; Adamczyk et al., 2009).

In moderate-speed walking, the potential energy (PE) and the kinetic energy (KE) are slightly out of phase (Figure 11.2; Saibene, 1990; Hallemans et al., 2004; Cavagna et al., 2005; Biewener, 2006; Grimshaw et al., 2007).

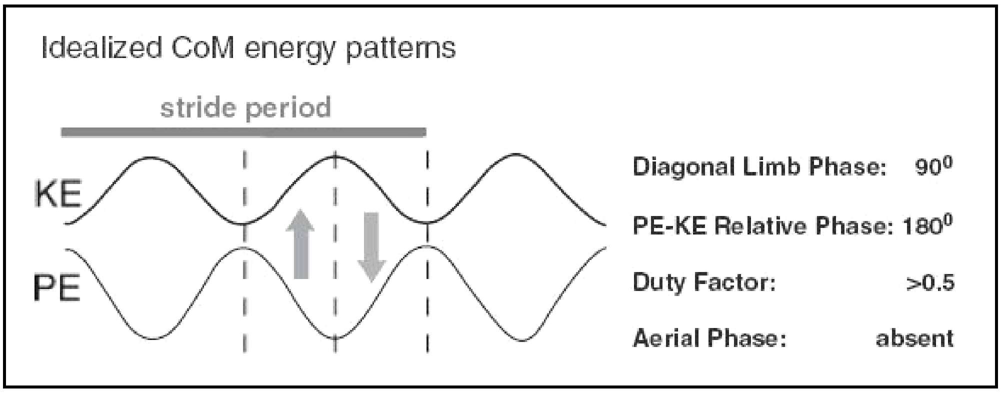


Figure 11.2. Kinetic energy (KE) and potential energy (PE) fluctuations of the BCOM during walking.

Specifically, in one walking cycle, both the speed of the BCOM and its height over the ground change (Zatsiorsky, 2002; Ortega et al., 2005; Biewener, 2006). Between touchdown and mid stance (see also chapter 1, par. 2.5.2), the forward velocity of the centre of mass decreases as the trunk arcs upwards over the stance foot. The highest position of the BCOM is at the mid stance. In this phase, kinetic energy is converted to gravitational potential energy (Zatsiorsky, 2002; Biewener, 2006). During the second phase, the centre of mass moves downwards as the forward velocity of the centre of mass increases. In this phase, potential energy is converted back into kinetic energy (Biewener et al., 2002; Ortega et al., 2005).

As a result, this exchange of potential energy and kinetic energy during walking is similar to the energy exchange of an oscillating pendulum. An ideal inverted pendulum system has perfect exchange between kinetic and potential energy (Cavagna et al., 1976; Mansour et al., 1982; Ortega et al., 2005). One reason why a human walker does not achieve 100% energy exchange is that the fluctuations in potential energy and kinetic energy are not matched in magnitude (Lee et al., 1998).

Therefore, the mathematical approach described in par. 1.2 above has been used to calculate KE, PE and TE during walking trials, in each age group (males and females).

In each graph below, examples of results concerning level walking (from 0 to 45 J, step 5 J) in our study are presented for all males (left graph) and females (right graph) aged 6 to 13 (Figure 11.3), 25 to 35 (Figure 11.4) and 56 to 65 (Figure 11.5).

Particularly, KE is illustrated in green, PE in red and TE in black.

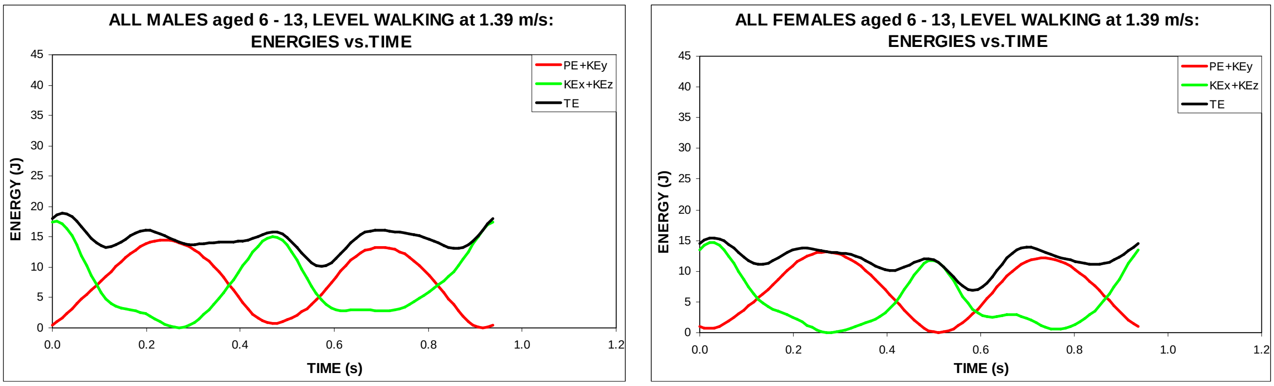


Figure 11.3. KE, PE and TE in level walking at 1.39 m/s,
males (on the left) and females (on the right) aged 6 to 13.

The pattern of kinetic, potential and total energy seems to be very similar in males (mean value = 14.665 J) and females (= 11.965 J) aged 6 to 13. Indeed, no qualitative significant differences are found.

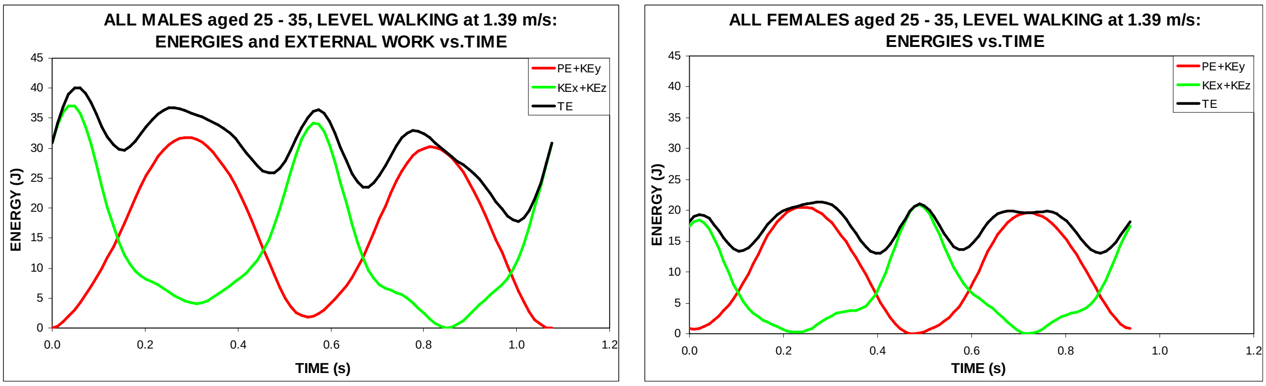


Figure 11.4. KE, PE and TE in level walking at 1.39 m/s,
males (on the left) and females (on the right) aged 25 to 35.

However, in males aged 25 to 35, both kinetic and potential energy are higher than corresponding values in females. Therefore, the total energy is always significantly greater in males

(= 30.196 J) than in females (= 17.426 J). This is probably due to both a higher forward velocity and a higher vertical displacement of the BCOM.

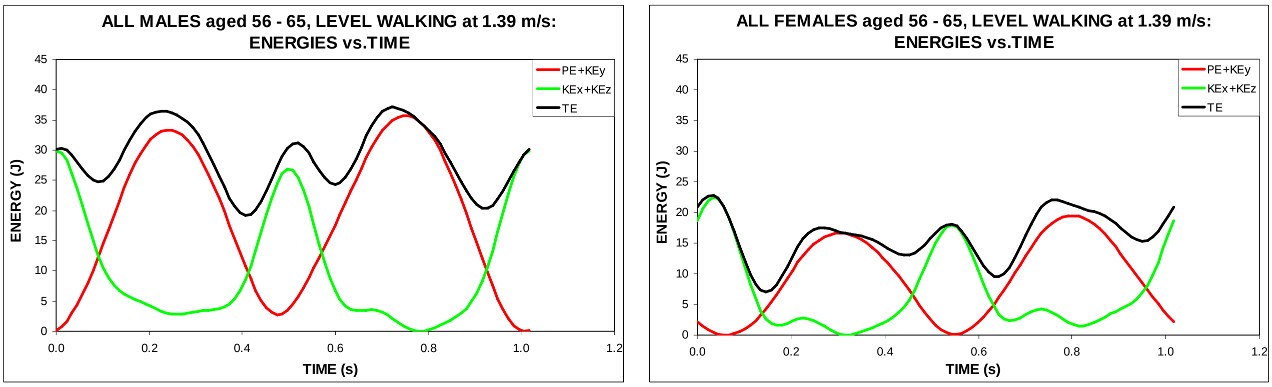


Figure 11.5. KE, PE and TE in level walking at 1.39 m/s,
males (on the left) and females (on the right) aged 56 to 65.

Similarly, in males aged 56 to 65, both kinetic and potential energy are higher than corresponding values in females. Therefore, the total energy is always significantly greater in males (= 28.796 J) than in females (= 16.105 J). As shown above, this is probably due to both a higher forward velocity and a higher vertical displacement of the BCOM.

As expected, all our results (including all testing walking speeds) concur with literature data (Cavagna et al., 1963; 1964; 1976; Cavagna, 1977a; Cavagna et al., 1977b; Gordon et al., 1980; Cavagna, 1988; Saibene, 1990; Minetti et al., 1993; Diedrich et al., 1995; Lee et al., 1998; Donelan et al., 2002a; Zatsiorsky, 2002; Robertson et al., 2004; Ortega et al., 2005; Biewener, 2006; Geyer et al., 2006; Ortega et al., 2007; Segers et al., 2007; Hof, 2008).

All these graphs in each age group are contained in the enclosed CD (First Study, Chapter 11, Level walking in males and females).

1.4. Potential and kinetic energy in running

As previously presented in chapter 1 (par. 4.4.2; Figure 1.11 and 1.12), kinetic (KE) and potential (PE) energy increases are in phase for running gaits (Figure 11.6; Cavagna et al., 2005; Biewener, 2006). However, this conceptual model should not be taken literally. Indeed, the **pogo-stick model** of running does not explain the mechanism of the forward propulsion (Zatsiorsky, 2002).

Precisely, as the foot hits the ground during running, the leg spring compresses as a result of flexion, and the mass moves downwards (Alexander, 1984; di Prampero, 1985; Cavagna, 1988; Blickhan, 1989; Saibene, 1990; Caldwell et al., 1992; Lee et al., 1998; Lejeune et al., 1998;

Alexander, 2005; Cavagna et al., 2005; Robilliard et al., 2005; Biewener, 2006; Cavagna et al., 2006; Hoyt et al., 2006; Bullimore et al., 2007; Segers et al., 2007; Grimmer et al., 2008).

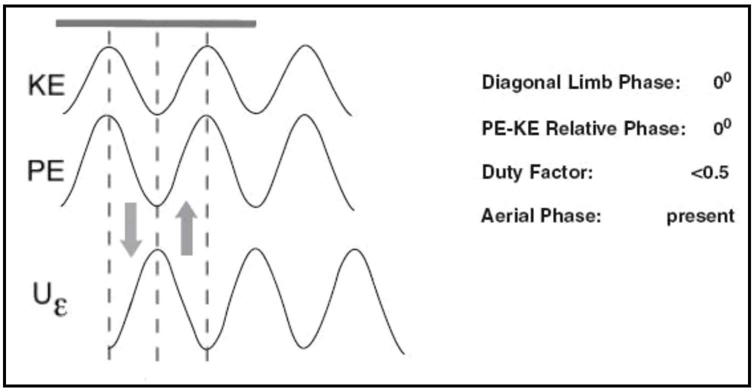


Figure 11.6. Kinetic energy (KE), potential energy (PE) and elastic energy (Uε) fluctuations of the BCOM during running.

Specifically, unlike walking, the BCOM is lowest at the mid stance. The maximal velocity and maximal position of the BCOM coincide in time (Zatsiorsky, 2002). Furthermore, KE and PE peak in mid-swing; when the foot comes into contact with the ground, KE is lost and when the BCOM falls to the ground, PE is lost. Much of the lost KE and PE is converted into elastic potential energy and stored in the muscles, tendons and ligaments (Cavagna, 1977a; Cavagna et al., 1977b; Luhtanen et al., 1978).

As in walking, in each graph below, examples of results concerning level running (from 0 to 160 J, step 20 J) in our study are presented for all males (left graph) and females (right graph) aged 6 to 13 (Figure 11.7), 25 to 35 (Figure 11.8) and 56 to 65 (Figure 11.9). Same colours already described in par. 1.3 have been used.

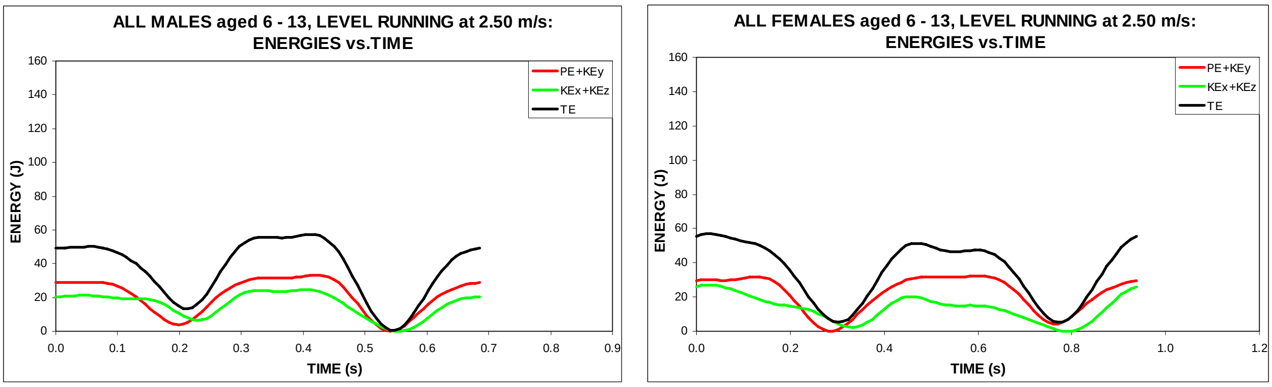


Figure 11.7. KE, PE and TE in level running at 2.50 m/s, males (on the left) and females (on the right) aged 6 to 13.

The pattern of kinetic, potential and total energy seems to be very similar in males (mean value = 36.999 J) and females (= 35.385 J) aged 6 to 13. Indeed, no qualitative significant differences are found.

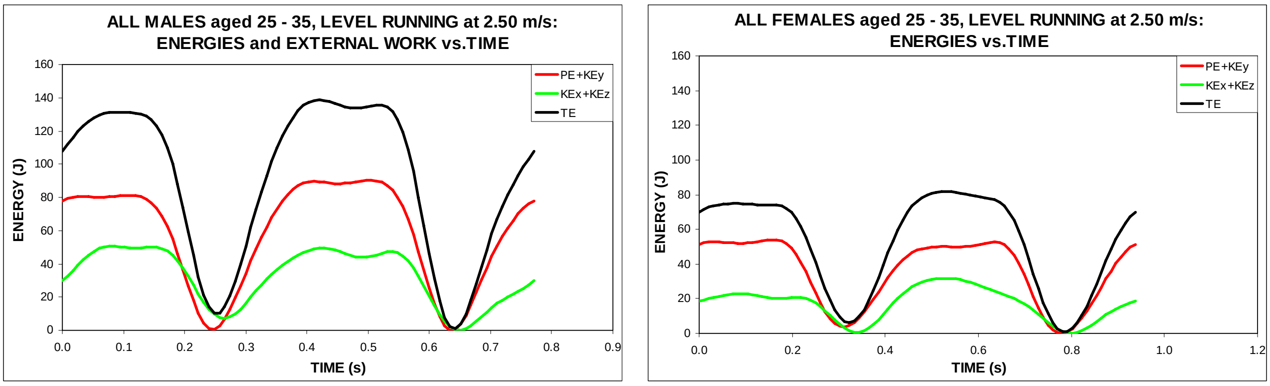


Figure 11.8. KE, PE and TE in level running at 2.50 m/s,
males (on the left) and females (on the right) aged 25 to 35.

However, in males aged 25 to 35, both kinetic and potential energy are higher than corresponding values in females. Therefore, the total energy is always significantly greater in males (= 91.236 J) than in females (= 52.655 J). As shown in walking, this is probably due to both a higher forward velocity and a higher vertical displacement of the BCOM.

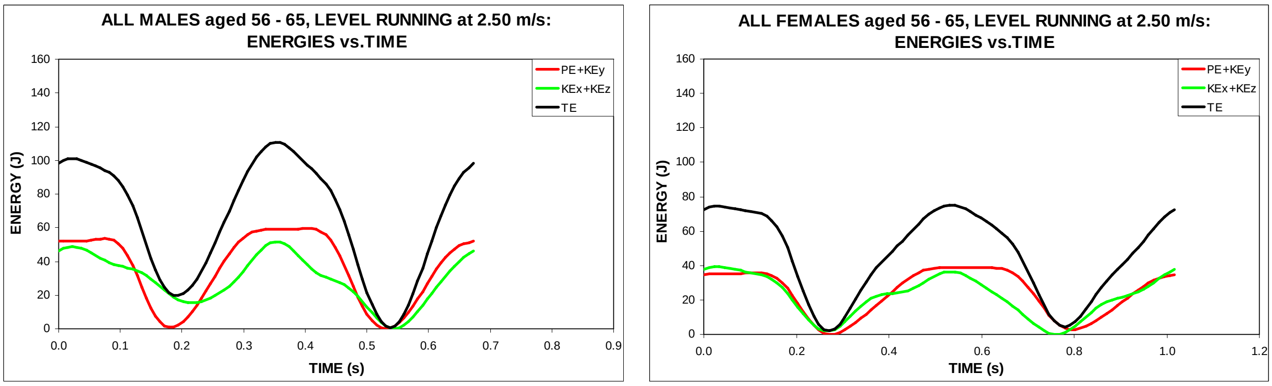


Figure 11.9. KE, PE and TE in level running at 2.50 m/s,
males (on the left) and females (on the right) aged 56 to 65.

Similarly, in males aged 56 to 65, both kinetic and potential energy are slightly higher than corresponding values in females. Therefore, the total energy is always greater in males (= 65.418 J) than in females (= 46.259 J). Clearly, this is probably due to both a higher forward velocity and a higher vertical displacement of the BCOM.

As expected, all our results (including all testing running speeds) concur with literature data (Cavagna et al., 1963; 1976; Cavagna, 1977a; Cavagna et al., 1977b; Gordon et al., 1980; Cavagna, 1988; Saibene, 1990; Caldwell et al., 1992; Minetti et al., 1993; Diedrich et al., 1995; Lee et al., 1998; Zatsiorsky, 2002; Robertson et al., 2004; Biewener, 2006; Cavagna et al., 2006; Geyer et al., 2006; Segers et al., 2007; Brughelli et al., 2008; Iida et al., 2008).

All these graphs in each age group are contained in the enclosed CD (First Study, Chapter 11, Level running in males and females).

2. STATISTICAL ANALYSIS

Mechanical external work (W_{ext}) and energy recovery percentage (R) have been calculated using Equation [1.17] and [1.18], respectively. In every testing condition, each of this variable has been graphically represented as shown in Figure 11.10 and 11.11, respectively.

Mechanical external work (J/(kg·m), blue) is represented as a function of time (s):

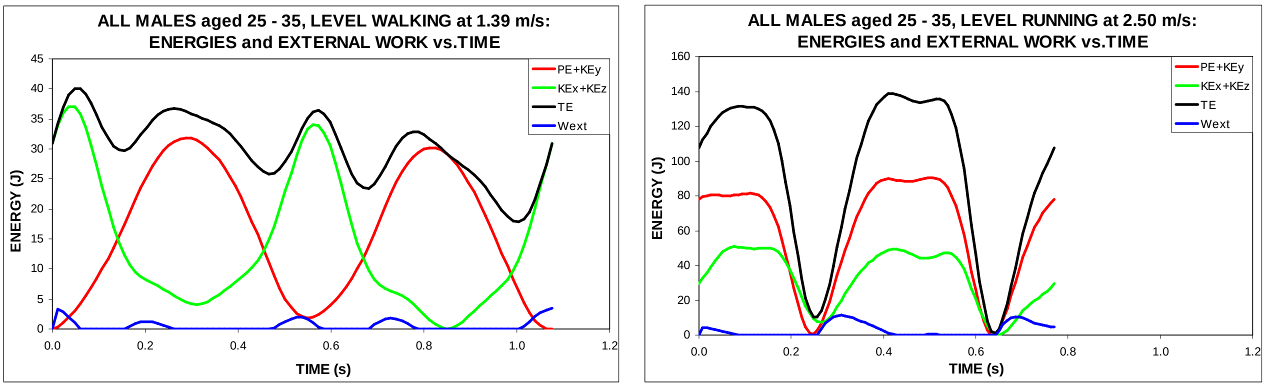


Figure 11.10. Mechanical external work in males aged 25 to 35,
level walking at 1.39 m/s (on the left) and level running at 2.50 m/s (on the right).

Energy recovery (as a fraction; pink) is represented as a function of time (s) or cycle by cycle:

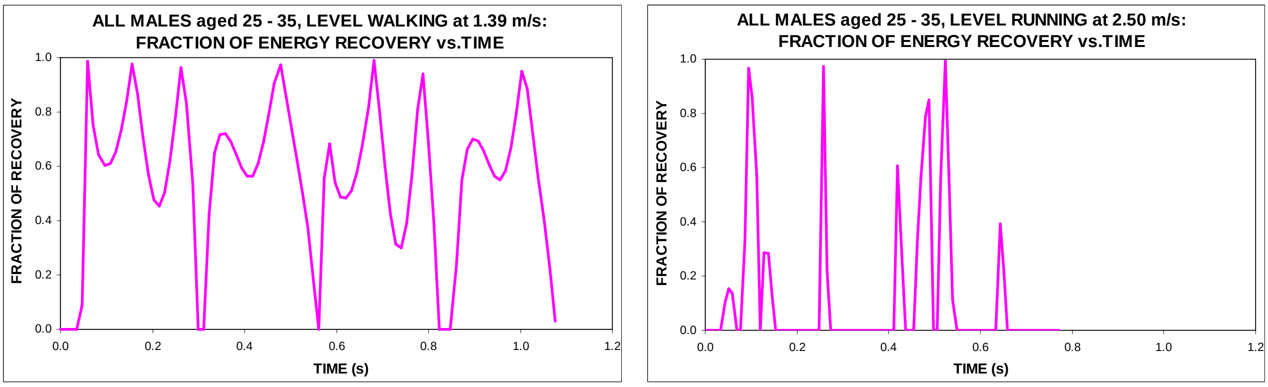


Figure 11.11. Fraction of energy recovery in males aged 25 to 35,
level walking at 1.39 m/s (on the left) and level running at 2.50 m/s (on the right).

Statistical analysis was performed by using average variable values.

Results will be presented as mean. The alpha test level set for statistical significance was 0.05.

The independent variables were age group (y) and progression speed (m/s). The chosen dependent variables were mechanical external work (W_{ext}) and energy recovery percentage (R).

Particularly, non-paired *t*-tests were performed to assess the effects between the dependent variable and each independent variable.

Effects of gender and age on each dependent variable were assessed by using a one-way ANOVA for unrelated measures. In addition, a post-*hoc* Bonferroni test was used to detect the strength of the associations between each dependent variable and gender/age. Moreover, effect of speed on each dependent variable was assessed by using a one-way ANOVA for related measures. In addition, a post-*hoc* paired *t*-test (with Bonferroni correction) was used to detect differences between each dependent variable and speed.

The graph legend is the same as already illustrated and described in chapter 8 (par. 3). Specifically, the average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 (see also chapter 6, par. 2.1) have been considered.

3. MECHANICAL EXTERNAL WORK

3.1. Mechanical external work as a function of age

A. In **level walking**, if W_{ext} is represented as a function of age (Figure 11.12), our results show that:

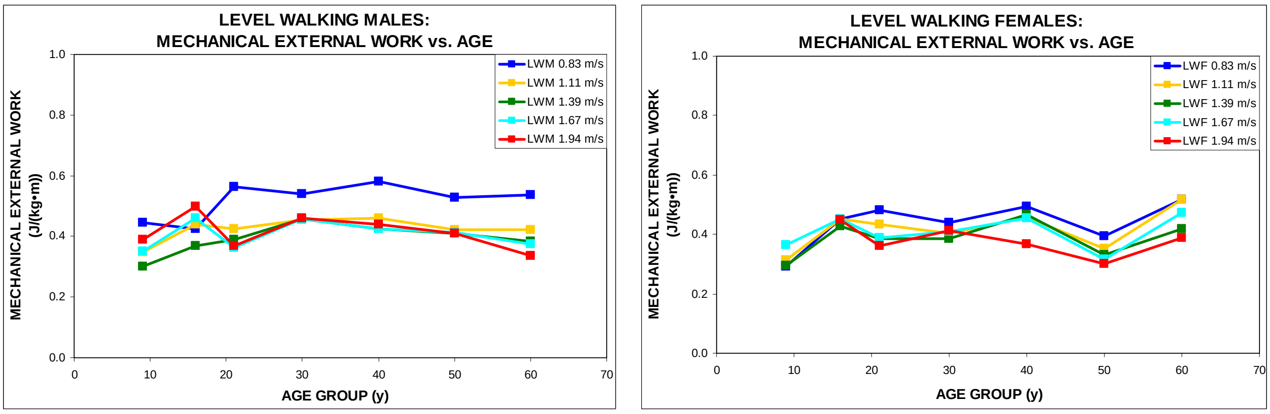


Figure 11.12. Mechanical external work as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), there is only little significant variability among all age groups (see Table 11.1 below). It is important to note that, above 18 years old, W_{ext} becomes significantly higher at the lowest speed of 0.83 m/s (0.550 ± 0.021 J/(kg·m), independently of age);

- in females (right graph), there is only little significant variability among all age groups (see Table 11.1 below);
- on average, W_{ext} values are similar both in males (0.444 ± 0.043 J/(kg·m), independently of age and speed) and females (0.412 ± 0.036 J/(kg·m));
- all these results partially confirm results obtained with the discrete method and previously discussed in chapter 10, par. 6.3.1. Indeed, by applying the continuous method, a slight underestimation could be observed in appreciating this complex biomechanical variable (i.e. probably due to the filtering procedure involved in calculating energies by using Fourier interpolation). Therefore, they partially concur with literature data (Cavagna et al., 1976; Schepens et al., 1998; 2001; Mian et al., 2006), too.

Specific results of the statistical analysis (with relevance) are shown in the table below.

LEVEL WALKING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	Males Females			$p<0.01$			
14 - 17 y		Males Females				$p<0.001$	
18 - 24 y			Males Females			$p<0.01$	
25 - 35 y	$p<0.01$			Males Females		$p<0.01$	
36 - 45 y					Males Females		
46 - 55 y		$p<0.001$	$p<0.01$	$p<0.01$		Males Females	
56 - 65 y							Males Females

Table 11.1. W_{ext} as a function of age in level walking (males and females).

B. In **level running**, if W_{ext} is represented as a function of age (Figure 11.13), our results show that:

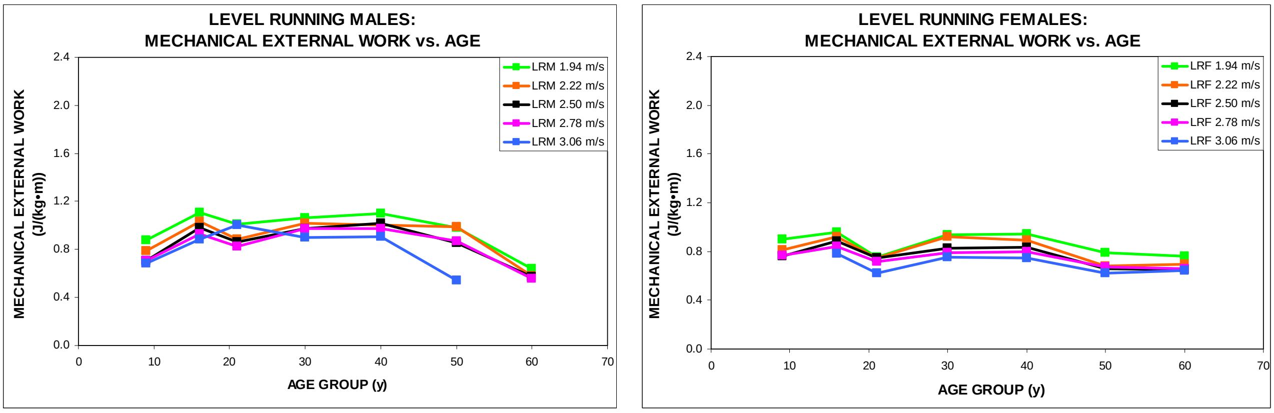


Figure 11.13. Mechanical external work as a function of age in level running, males (on the left) and females (on the right).

- in males (left graph), there is little significant variability among all age groups (see Table 11.2 below). Indeed, on average, adult males aged 56 to 65 have a lower value of W_{ext} (0.579 ± 0.036 J/(kg·m), independently of speed) which is significant different from the other age groups (0.923 ± 0.097 J/(kg·m); $p<0.001$). Importantly, this pattern is quite similar at each speed;
- in females (right graph), there is little significant variability among all age groups (see Table 11.2 below);
- all these results do not confirm results obtained with the discrete method and previously discussed in chapter 10, par. 6.5.1. Indeed, by applying the continuous method, a significant underestimation could be observed in appreciating this complex biomechanical variable (i.e. probably due to the filtering procedure involved in calculating energies by using Fourier interpolation). Yet, they partially concur with literature data (Cavagna et al., 1976; 2008).

Specific results of the statistical analysis (with relevance) are shown in the table below.

LEVEL RUNNING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	**Males** **Females**	$p<0.001$		$p<0.001$	$p<0.001$		$p<0.01$
14 - 17 y	$p<0.001$	**Males** **Females**	$p<0.01$			$p<0.001$	$p<0.001$ $p<0.001$
18 - 24 y		$p<0.01$	**Males** **Females**	$p<0.05$			$p<0.001$
25 - 35 y	$p<0.001$		$p<0.05$	**Males** **Females**		$p<0.01$	$p<0.001$ $p<0.01$
36 - 45 y	$p<0.001$				**Males** **Females**	$p<0.01$	$p<0.001$ $p<0.01$
46 - 55 y		$p<0.001$		$p<0.01$	$p<0.01$	**Males** **Females**	$p<0.001$
56 - 65 y	$p<0.01$	$p<0.001$ $p<0.001$	$p<0.001$	$p<0.001$ $p<0.01$	$p<0.001$ $p<0.01$	$p<0.001$	**Males** **Females**

Table 11.2. W_{ext} as a function of age in level running (males and females).

3.2. Mechanical external work as a function of speed

A. In **level walking**, if W_{ext} is represented as a function of speed (Figure 11.14), our results show that:

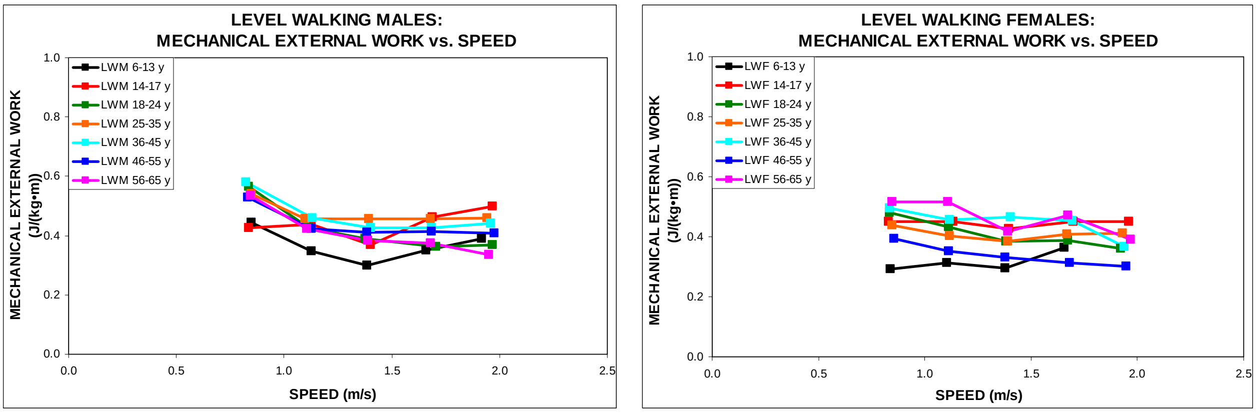


Figure 11.14. Mechanical external work as a function of speed in level walking, males (on the left) and females (on the right).

- in males (left graph), W_{ext} decreases from 0.83 to 1.39 m/s (from 0.517 ± 0.058 to 0.390 ± 0.050 J/(kg·m), $p<0.001$), independently of age; above this speed, it does not significantly change. Importantly, this pattern is quite similar in each age group;
- in females (right graph), W_{ext} does not significantly change as walking speed changes. Importantly, this pattern is quite similar in each age group. Indeed, it is important to underline the variable pattern in females aged 56 to 65: W_{ext} decreases from 0.83 to 1.11 m/s (from 0.536 to 0.421 J/(kg·m), $p<0.001$); it then increases from 1.11 to 1.67 m/s (from 0.421 to 0.476 J/(kg·m), $p<0.01$); finally, it decreases from 1.67 to 1.94 m/s (from 0.476 to 0.336 J/(kg·m), $p<0.001$);
- all these results do not completely confirm results obtained with the discrete method and previously discussed in chapter 10, par. 6.3.2. However, they concur with some literature data (Ardigò, 1992; Willems et al., 1995; Saibene et al., 2003; Mian et al., 2006).

Specific results of the statistical analysis (with relevance) are shown in the table below.

SUBJECTS	**LEVEL WALKING in each age group**
Males	$p<0.001$ from 0.83 to 1.39 m/s and $p=NS$ from 1.39 to 1.94 m/s
Females	$p=NS$ comparing all speeds

Table 11.3. W_{ext} as a function of speed in level walking (males and females).

B. In **level running**, if W_{ext} is represented as a function of speed (Figure 11.15), our results show that:

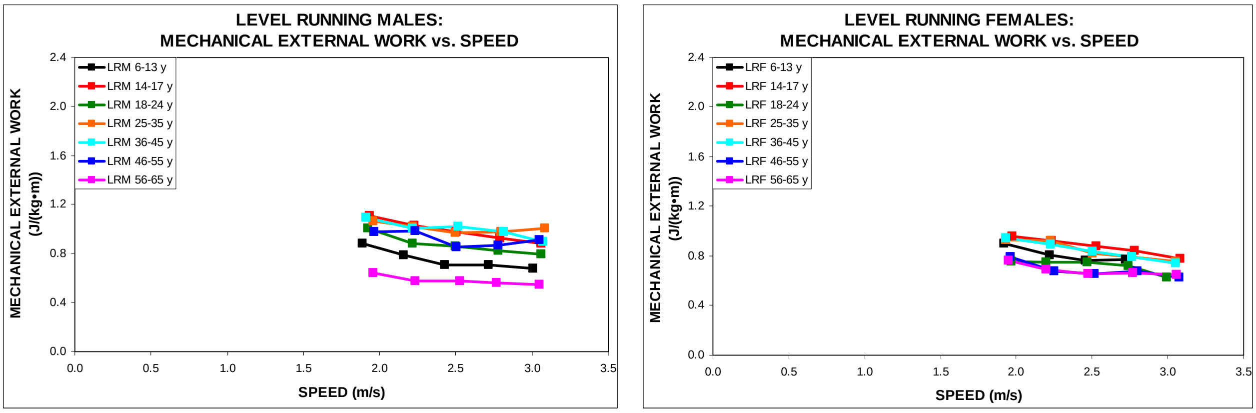


Figure 11.15. Mechanical external work as a function of speed in level running, males (on the left) and females (on the right).

- both in males (left graph) and females (right graph), W_{ext} does not change as running speed changes. Importantly, this pattern is quite similar in each age group;
- all these results do not confirm results obtained with the discrete method and previously discussed in chapter 10, par. 6.4.2. However, an absence of the relationship with speed has been satisfied, as well. Finally, all these results concur with some literature data (Ardigò, 1992; Willems et al., 1995; Saibene et al., 2003; Mian et al., 2006).

Specific results of this statistical analysis (with relevance) are shown in the table below.

SUBJECTS	LEVEL RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 11.4. W_{ext} as a function of speed in level running (males and females).

4. ENERGY RECOVERY PERCENTAGE

4.1. Energy recovery percentage as a function of age

In **level walking**, if R is represented as a function of age (Figure 11.16), our results show that:

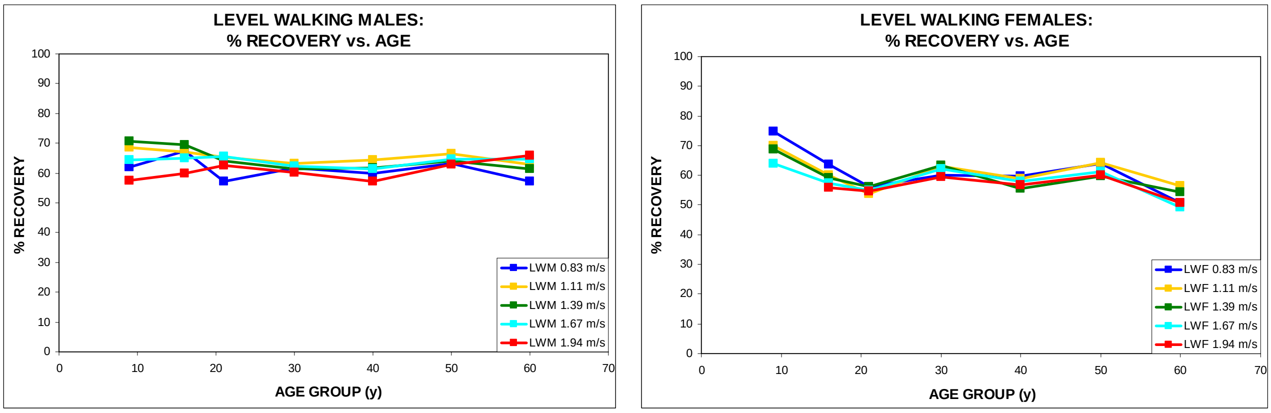


Figure 11.16. Energy recovery percentage as a function of age in level walking, males (on the left) and females (on the right).

- in males (left graph), energy recovery is slightly greater in young children aged 6 to 13 (64.592 ± 3.950%, at all the investigated speeds); however, there are no significant differences in the other age groups (62.989 ± 1.744%). Finally, this pattern is similar at all speeds;
- in females (right graph), energy recovery is greater in young children aged 6 to 13 (57.569 ± 4.507%, at all the investigated speeds); however, there are only slight significant differences in the other age groups. Importantly, this pattern is similar at all speeds;
- both in males and females, values of energy recovery obtained by applying the continuous method are not completely similar to values obtained by applying the discrete method. As a consequence, all these results partially confirm results obtained with the discrete method and previously discussed in chapter 10, par. 7.2.1. Indeed, by applying the continuous method, a slight underestimation could be observed in appreciating this complex biomechanical variable (i.e. probably due to the filtering procedure involved in calculating energies by using Fourier interpolation). We could conclude that all these results partially concur with literature data (Cavagna et al., 1976; 1983).

Specific results of the statistical analysis (with relevance) are shown in the table below.

LEVEL WALKING	6 - 13 y	14 - 17 y	18 - 24 y	25 - 35 y	36 - 45 y	46 - 55 y	56 - 65 y
6 - 13 y	Males Females						
14 - 17 y		Males Females	p<0.01				
18 - 24 y		p<0.01	Males Females	p<0.001		p<0.001	
25 - 35 y			p<0.001	Males Females	p<0.01	p<0.01	
36 - 45 y				p<0.01	Males Females	p<0.05 p<0.01	
46 - 55 y			p<0.001	p<0.01	p<0.05 p<0.01	Males Females	
56 - 65 y							Males Females

Table 11.5. R as a function of age in level walking (males and females).

4.2. Energy recovery percentage as a function of speed

In **level walking**, if R is represented as a function of speed (Figure 11.17), our results show that:

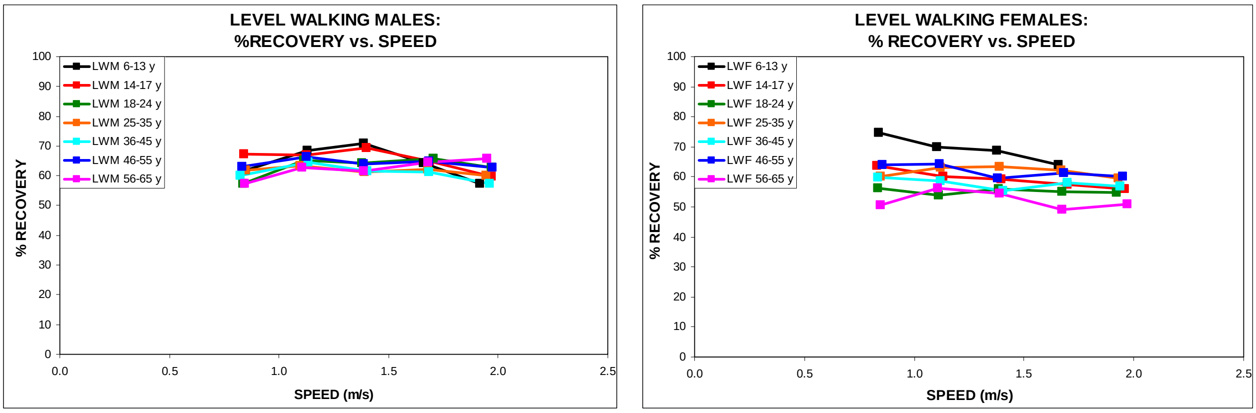


Figure 11.17. Energy recovery percentage as a function of speed in level walking, males (on the left) and females (on the right).

- the maximum recovery speed increases the older the subject, or the 'optimal speed' of walking is the smallest the younger the subject (Cavagna et al., 1983; Willems et al., 1995);
- indeed, our results show that this optimal speed ranges from ≈ 1.11 to ≈ 1.39 m/s in the youngest subjects aged 6 to 13, and it is higher than ≈ 1.39 m/s in young and elderly adults aged 14 to 65. Above this 'optimal speed', the energy recovery decreases (Cavagna et al., 1983): this is more evident in children (both in males and females, aged 6 to 17), and in elderly females (aged 56 to 65);
- all these results do not completely confirm results obtained with the discrete method and previously discussed in chapter 10, par. 7.2.2. Furthermore, they partially concur with some literature data (Cavagna et al., 1983; Minetti et al., 1995; Willems et al., 1995; Mian et al., 2006).

Specific results of this statistical analysis (with relevance) are shown in the table below.

LEVEL WALKING	0.83 m/s	1.11 m/s	1.39 m/s	1.67 m/s	1.94 m/s
0.83 m/s	**Males** **Females**				
1.11 m/s		**Males** **Females**			$p<0.01$ $p<0.05$
1.39 m/s			**Males** **Females**		
1.67 m/s				**Males** **Females**	$p<0.05$
1.94 m/s		$p<0.01$ $p<0.05$		$p<0.05$	**Males** **Females**

Table 11.6. R as a function of speed in level walking (males and females).

5. CONTINUOUS ANALYSIS *VERSUS* DISCRETE ANALYSIS

5.1. Introduction

In this last section, we will compare results obtained with the two methods (Bland et al., 1986):

1. **discrete analysis**, based upon the study of cycle by cycle pattern in each biomechanical variable. For more details about results obtained applying this method, see also chapter 10;
2. **continuous analysis**, based upon the study of the harmonics pattern in order to investigate each biomechanical variable. For more details about this mathematical method, see also chapter 6.

For this reason, mechanical external work (W_{ext}) and energy recovery percentage (R) calculated with mathematical method are compared to the corresponding values obtained with cycle by cycle method and *vice versa*. This comparison states and verifies the similarities and/or the differences among the methods. In this way, we answer the crucial question: '*could the continuous analysis take the place of the discrete method because of the homogeneity of results?*'.

Average values of each comparison are contained in the CD (First Study, Chapter 11, Comparison between the two methods).

5.2. About mechanical external work …

A. In **level walking** (Figure 11.18), our results show that:

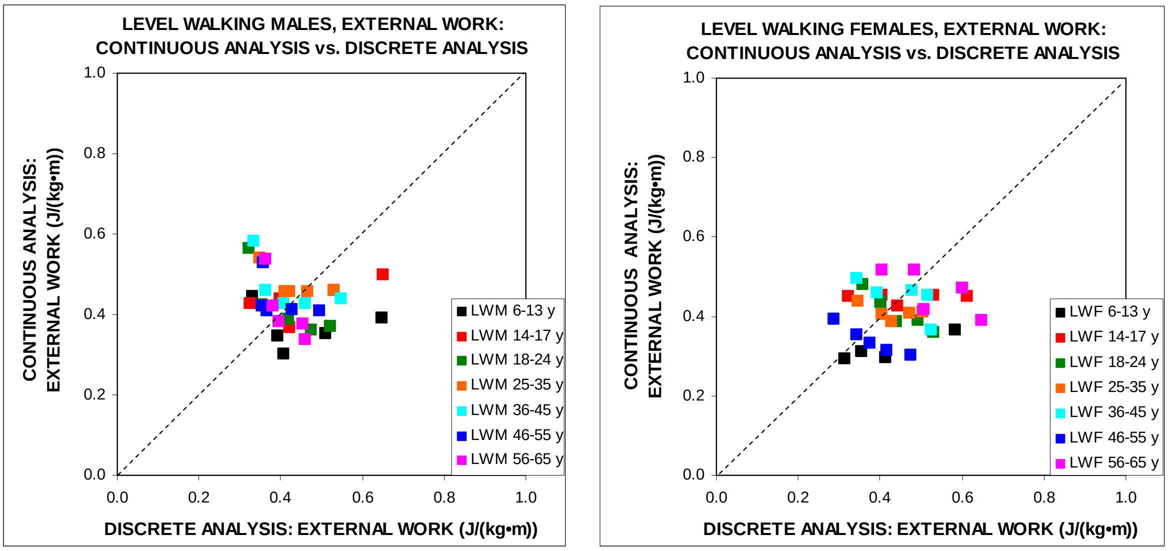


Figure 11.18. Mechanical external work (continuous analysis) *versus* mechanical external work (discrete analysis) in level walking, males (on the left) and females (on the right).

- both in males (left graph; n = 70, R^2 = 0.426, r = 0.620) and females (right graph; n = 70, R^2 = 0.559, r = 0.683), independently of speed, there are only slight significant differences between the discrete analysis and the continuous analysis (see also the location of the data with respect to the identity line). Importantly, this quite similarity occurs in each age group;
- as a result, we can conclude that the two methods could be interchangeable. Furthermore, it is important to observe: 1) a wider horizontal scattering in results derived by the discrete analysis; and 2) a closer vertical scattering in results derived by the continuous analysis.

Finally, in this case, the answer to the initial question '*could the continuous analysis take the place of the discrete method because of the homogeneity of results?*' is quite positive.

B. In **level running** (Figure 11.19), our results show that:

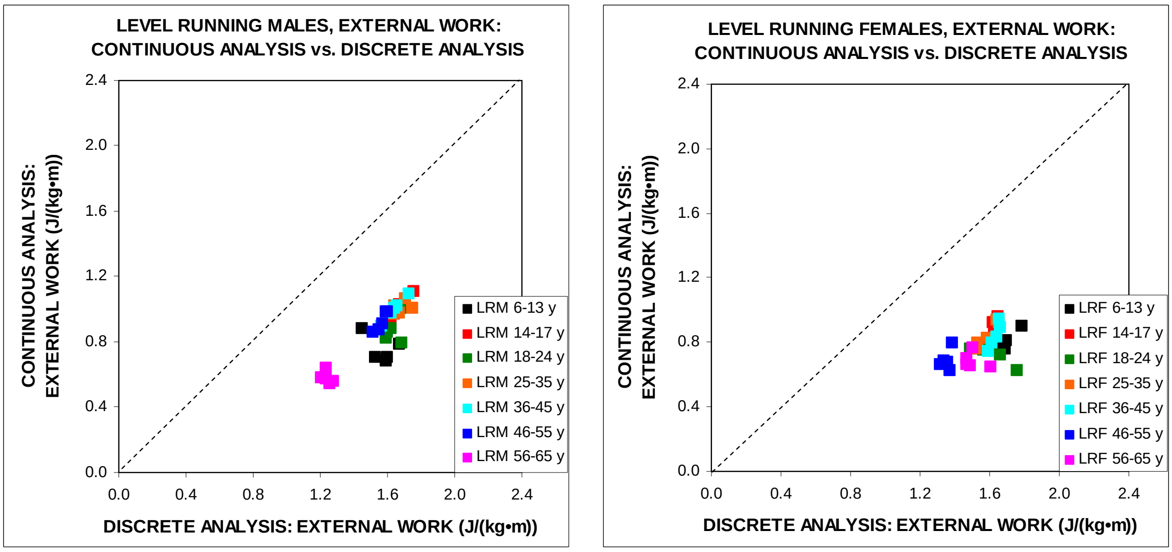


Figure 11.19. Mechanical external work (continuous analysis) *versus* mechanical external work (discrete analysis) in level running, males (on the left) and females (on the right).

- both in males (left graph; n = 70, R^2 = 0.509, r = 0.660) and in females (right graph; n = 70, R^2 = 0.637, r = 0.758), independently of speed, there are significant differences between the discrete analysis and the continuous analysis ($p<0.001$);
- importantly, as previously reported, external work is always higher when it is calculated applying the discrete method (see also the location of the data with respect to the identity line: they are in the right panel). This important result is probably due to the filtering procedure involved in calculating energies by using Fourier coefficients. Therefore, the continuous method significantly underestimates results;
- importantly, this pattern is quite similar in each age group;
- as a result, these evident differences allow us to conclude that the two methods could not be interchangeable. Furthermore, it is important to observe: 1) a wider horizontal scattering in results derived by the discrete analysis; and 2) a closer vertical scattering in results derived by the continuous analysis. In addition, this pattern seems to be more evident in females.

Finally, in this case, the answer to the initial question ‘*could the continuous analysis take the place of the discrete method because of the homogeneity of results?*’ is absolutely negative.

5.3. About energy recovery percentage …

In **level walking** (Figure 11.20), our results show that:

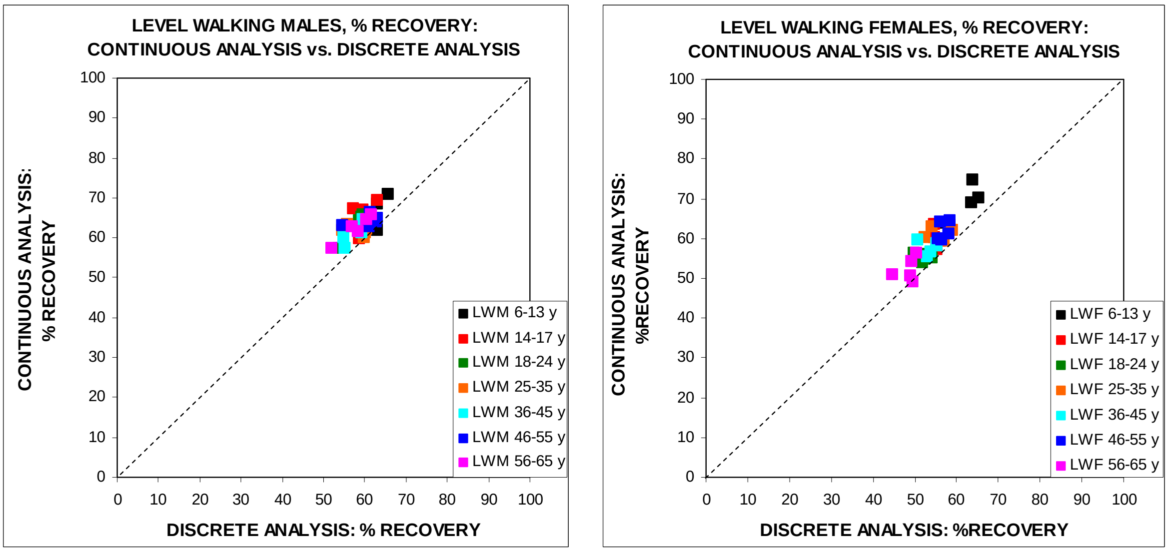


Figure 11.20. Energy recovery percentage (continuous analysis) *versus* energy recovery percentage (discrete analysis) in level walking, males (on the left) and females (on the right).

- both in males (left graph; n = 70, $R^2 = 0.552$, r = 0.702) and females (right graph; n = 70, $R^2 = 0.196$, r = 0.399), independently of speed, there are only slightly significant differences ($p<0.05$) between the discrete analysis and the continuous analysis. This pattern is similar in each age group;
- importantly, energy recovery is slightly higher when it is calculated with the continuous method (see also the location of the data with respect to the identity line: they are in the left panel). This result is probably due to the filtering procedure involved in calculating energies by using Fourier coefficients, as well. Therefore, the discrete method underestimates results;
- as a result, we can conclude that the two methods could not be perfectly interchangeable. Furthermore, it is important to observe: 1) a wider horizontal scattering in results derived by the discrete analysis; and 2) a closer vertical scattering in results derived by the continuous analysis. This pattern seems to be more evident in females.

Finally, in this case, the answer to the initial question '*could the continuous analysis take the place of the discrete method because of the homogeneity of results?*' is quite negative.

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Chapter 12

VALIDATION OF A MODEL EQUATION TO PREDICT INTERNAL WORK

1. INTRODUCTION

Analysis of the motion of the body during locomotion is of great interest to many biological disciplines (e.g. Physiology, Physics, Biomechanics and so on). Therefore, a common aim within the study of the mechanics of human locomotion is the calculation of the mechanical work performed (Winter, 1978; Caldwell et al., 1992; Winter, 2005; Mahaudens et al., 2009).

As already described in chapter 10, the mechanical total work of locomotion (W_{tot}) has been traditionally regarded as the sum of mechanical external work (W_{ext}; see also chapter 10, par. 6) and mechanical internal work (W_{int}; see also chapter 10, par. 8), which are considered two separate entities (Cavagna et al., 1976; 1977; Saibene et al., 2003; Winter, 2005).

On one hand, the **mechanical external work** represents the work necessary to lift and accelerate the BCOM within the environment (Aleshinsky, 1986a; Lejeune et al., 1998; Sparrow, 2000; Cerretelli, 2001; Saibene et al., 2003; Minetti, 2004; Mahaudens et al., 2009). It has been investigated in many different conditions and populations (Cavagna et al., 1983; Minetti et al., 1994; Schepens et al., 1998; 2001; Kuo et al., 2005; Sasaki et al., 2008). From the above definition, W_{ext} requires the potential energy (PE) and the kinetic energy (KE) of the BCOM to be measured; furthermore, the total energy (TE = PE + KE) and its change over time need to be calculated (Cavagna et al., 1976; 1991; Willems et al., 1995; Duff-Raffaele et al., 1996; Saibene et al., 2003; Minetti, 2004). This goal can be achieved both by using dynamometric (direct dynamics) and motion analysis (inverse dynamics) technique (Avogadro et al., 2003; Purkiss et al., 2003; Robertson et al., 2004; Richards, 2008; Sasaki et al., 2008).

On the other hand, the movements of the body segments relative to the BCOM are to a large extent (but not exclusively) brought about by forces internal to the body and, consequently, the **mechanical internal work**, associated with the mechanical energy changes relative to the BCOM, constitutes the work necessary a) to accelerate the limbs with respect to the BCOM during human locomotion (Cavagna et al., 1964; 1977; Winter, 1979; Aleshinsky, 1986a; 1986b; Steudel, 1990; Caldwell et al., 1992; Minetti et al., 1992; 1993; 1994; Aissaoui et al., 1996; Lejeune et al., 1998; Minetti et al., 1998; Schepens et al., 1998; Minetti et al., 1999; Sparrow, 2000; Cerretelli, 2001; Schepens et al., 2001; Purkiss et al., 2003; Saibene et al., 2003; Hallemans et al., 2004; Minetti, 2004; Winter, 2005; Cavagna et al., 2008; Sasaki et al., 2008; Mahaudens et al., 2009), b) to overcome internal friction or viscosity (Minetti et al., 2006), c) to overcome antagonistic co-contractions, and d) to stretch the series elastic components (Bastien et al., 2003). It is usually computed from both reciprocal segment movements (Sasaki et al., 2008) and anthropometric parameters (Schepens et al., 2001). Furthermore, W_{int} constitutes 25-40% of the total mechanical work in humans during locomotion (Aissaoui et al., 1996; Duff-Raffaele et al., 1996; Bastien et al., 2003; Saibene et al., 2003).

2. MECHANICAL INTERNAL WORK CALCULATION: A BRIEF HISTORY

2.1. Introduction

Historically, the concept of W_{int} was introduced by Fenn (for running at top speeds, 1930) (Cavagna et al., 1964; 1976; Minetti, 2004); it was then formalized by Cavagna and Kaneko (Cavagna et al., 1977).

Particularly, the mechanical internal work (W_{int}) derives from the König theorem of physics (Cavagna et al., 1986). This theorem stated that '*the kinetic energy of a system of particles is the kinetic energy associated to the movement of the center of mass and the kinetic energy associated to the movement of the particles relative to the center of mass*' (Cavagna et al., 1986). Moreover, it verified that, in a linked multi-segment system, '*the total kinetic energy can be partitioned in two different components: first of all, that of the BCOM with respect to the environment*' (the so-called W_{ext}), and '*secondly, that of single segments with respect to the BCOM*' (the so-called W_{int}).

Consequently, the biomechanical interest in W_{int} resides in the capability to consider the acceleration of body segments whose movements do not directly result in a change for the BCOM position (Cavagna et al., 1964; Winter, 1978; Aleshinsky, 1986b; Minetti, 1998) which is the case of human locomotion (walking and running, in particular), where limbs are moved quasi-symmetrically with respect to the BCOM (Minetti et al., 1993).

2.2. Historical internal work calculation

In general, calculation of W_{int} is more complicated than of W_{ext}. Thus, the recordings of the mechanical energy level of the individual body segments, obtained by cinematography, are far 'more complex and difficult to interpret' (Willems et al., 1995). Furthermore, calculation of W_{int} requires assumptions about the physical properties of the body segments, as well as regarding the transfer of energy to and from different body segments. Indeed, W_{int} can be calculated by summing the kinetic energy curves of single segments (in a way which allows energy transfer only among within-limb segments) and by summing all up the energy increases of the resulting curves (Fenn, 1930; Willems et al., 1995).

Different computational models have been proposed to calculate the mechanical internal work. Most of these models use the traditional approach of examining changes in segmental energies (*absolute work method*; Winter, 1978; Purkiss et al., 2003); otherwise, most use inverse dynamics and joint power analysis (*absolute power method*; Ferris et al., 1998; Purkiss et al., 2003). To be precise, the absolute power method is considered to be 'the superior technique for quantifying mechanical internal work in elite runners' (Purkiss et al., 2003).

The various models/techniques for calculating the mechanical internal work (in the field of human gait) have undergone a general improvement over the years (Table 12.1).

TECHNIQUE	WORK COMPONENTS NOT ACCOUNTED FOR BY TECHNIQUE		
Increase PE or KE Fenn (1930) Saunders et al. (1953) Liberson (1965)	Energy exchanges within segments and transfers between segments	Simultaneous increases or decreases in reciprocally moving segments	Simultaneous generation and absorption in different joints
Centre of mass Cavagna et Margaria (1968)		Simultaneous increases or decreases in reciprocally moving segments	Simultaneous generation and absorption in different joints
∑ segments energies Winter (1979)			Simultaneous generation and absorption in different joints
Joint power Winter (1953)			
Muscle power Yack (1985)			

Table 12.1. Techniques to calculate internal work in movement, in Winter (2005).

Fenn (1930) summed the increases of energy in each of the major segments, over the stride period, to yield the net mechanical work (Lloyd et al., 1972). His hypothesis was that '*the kinetic energy turns out to be high in that limb where the work is being done. If the kinetic energy is calculated in relation to the ground, then the limb going backwards has very small kinetic energy although the actual effort on the part of the runner is as great in pushing it backwards as in pushing it forwards*'. Indeed, he did consider both energy exchanges within segments and passive transfers between segments (Cavagna et al., 1977; 1988).

Cavagna and Margaria (1966) proposed a technique based on the potential and kinetic energies of the BCOM. They recorded these energies by a force platform during both walking and running. Yet, they made the erroneous assumption that the BCOM reflects the energy changes in all segments.

Ralston and Lukin (1969) and Winter (1979) calculated the kinetic and potential energies of the major segments by means of displacement transducers and TV imaging techniques. Unfortunately, their calculation underestimated the simultaneous energy generation and absorption at different joints (Winter, 1979).

Cavagna (1977) determined the internal work in relation to the velocity of the shoulder joint for both the arm and the hip joint in the leg, with the assumption that these joints do not move relatively to the BCOM during locomotion. However, BCOM is expected to move more 'smoothly' than those joints and their kinetic energy might change less than the amount obtained with that methodology.

Finally, Winter calculated the internal work from the sum of segment energies which he then compared to the same calculation on the BCOM energy (Winter, 1979; 2005).

2.3. Direct measurement *versus* indirect measurement

Since its biomechanical definition, W_{int} has been widely investigated in literature. As a result, it seems to be useful in comparative and intra-species analysis of mechanical relationships during human locomotion in different conditions (Cavagna et al., 1997), gaits (Minetti et al., 1993), gradients (Minetti et al., 1993) and stride frequencies (Cavagna et al., 1983; 1986; 1988; 1991; Minetti et al., 1992; Cavagna et al., 1997).

Despite its scientific relevance, only a few laboratories measure this variable, because of methodological limitations. Differently to W_{ext}, W_{int} categorically requires the expensive cinematographic method so that an experimental set-up and a complex computer program *ad hoc* are essential (Minetti, 1998). As a consequence, nowadays very little data from direct measurements (inverse dynamics) exists of the mechanical internal work actually done by the muscles in human exercises such as walking and running.

Consequently, some researchers have tried to define a mathematical method to evaluate this biomechanical variable when the direct measurement is not available.

Among the others, the need to refer to a standard equation for W_{int} has been solved by Minetti (1998). He has provided a general mathematical equation to estimate the mechanical internal work (PW_{int}) in human locomotion at different gaits, speeds, frequencies and gradients. As a consequence, in order to better understand and explain results already presented/discussed in chapter 10 (referring to the Measured mechanical internal work MW_{int}; see par. 8.2), we used this equation proposed by Minetti to calculate the Predicted mechanical internal work (see also Avogadro et al. (2003) and Formenti et al. (2005) who applied the same mathematical method or the indirect measurement).

Specifically, the Predicted mechanical internal work (PW_{int}) has been calculated for both human walking and running in the different age groups moving at different speeds and gradients (according to the subjects and protocol test previously described in chapter 5). As a result, the Measured internal work (MW_{int}) was compared to the Predicted internal work (PW_{int}). Of utmost significance, these same measurements were used to specifically investigate how these variables change with all testing conditions (gender, age, type of locomotion, speed and gradient).

3. THE MODEL EQUATION FOR THE PREDICTION OF INTERNAL WORK

3.1. Introduction

By refining a previously published model, Minetti (1998) proposed a simple equation for the estimation of the mechanical internal work during human locomotion (see also Appendix, at the end of this chapter). As a result, the Predicted mechanical internal work expressed as J *per* kg of body mass *per* unit distance travelled (m), a customary unit for the mechanical and metabolic cost of locomotion, can be expressed as:

$$PW_{int} = SF \bullet s \bullet (1 + (\frac{DF}{1 - DF})^2) \bullet q \quad [\text{Eq. } 12.1]$$

where *SF* is the stride frequency (see also chapter 10, par. 2); *s* is the average progression speed; *DF* is the duty factor (see also chapter 10, par. 5); and *q* is a compound dimensionless term accounting for the inertial properties of the limbs and the mass partitioned between the limbs and the rest of the body (see par. 3.2 below).

In humans (bipeds), fore and hind limbs are alternatively in contact with the ground, while the upper limbs oscillate freely both during the stance and the swing phase. For the mathematical model works properly, it has been assumed that the duty factor to put in the equation will be the same both in upper and lower limbs (see Appendix). This approximation is mitigated by the lower mass and the inertial moment of upper limbs with respect to lower limbs (Dempster et al., 1959).

The practical meaning of the standard Equation [12.1] is that W_{int} can be predicted by knowing: a) the stride frequency; b) the average progression speed; c) the duty factor; and d) the mean value of *q* for the subjects under investigation, regardless of the gait type.

3.2. The compound dimensionless term *q*

3.2.1. General mathematical definition

The main problem in the previous model is the mathematical definition of the term *q*, which could be expressed as (Minetti et al., 1992):

$$q = \left(\frac{\pi^2}{4}\right) \bullet [(a^2 + g^2) \bullet (ml^* + b^2 \bullet mu^*)] \quad \text{[Eq. 12.2]}$$

where *π* is a constant; *a* is the fractional distance of the lower limb centre of mass to the proximal joint; *g* is the average radius gyration of limbs, as a fraction of the limb length; *b* is the length of the upper limb, as a fraction of the lower limb; with *mu** and *ml** being the fractional mass of the upper and lower limb mass, respectively.

Values of single variables defining *q* are shown in Table 12.2. As already illustrated, anthropometric parameters could be referred to Winter (2005) for males, and to both Winter (2005) and Dempster et al. (1959), Zatsiorsky et al. (1990), de Leva et al. (1996) for females (see also chapter 6, par. 2.1).

	MALES	**FEMALES**
a	0.447	0.447
g	0.378	0.399
b	0.440/0.491	0.440/0.491
mu*	0.050	0.045
ml*	0.161	0.170

Table 12.2. Single components of *q* in order to measure the Predicted mechanical internal work.

As a result and as stated in the PW_{int} equation, *q* values reflect the inertial properties of the limbs. Indeed, *a* and *g* can be changed by the degree of flexion of the different segments composing the limb during the swing phase (Minetti, 1998).

3.2.2. Practical mathematical definition

Equation [12.3] shows that, by using Equation [12.1], the term *q* could also be calculated as:

$$q = \frac{MW_{int}}{SF \bullet s \bullet \left(1 + \left(\frac{DF}{1 - DF}\right)^2\right)} \quad \text{[Eq. 12.3]}$$

where MW_{int} is the Measured internal work (experimental average internal work, derived from kinematic data); *SF* is the experimental average stride frequency; *s* is the average speed progression; and *DF* is the average term related to the duty factor.

The main values of all these parameters have been processed according to Equation [12.3] in order to obtain the term *q*. The term *q* has been therefore calculated to verify if it varies in relation to gender, age, speed and gradient.

Then, the average q (= 0.08 at level gaits and = 0.10 at gradients gaits; see par. 4.5 below) has been computed and fed back into Equation [12.1] to predict the mechanical internal work (PW_{int}).

4. THE TERM *q* IN OUR EXPERIMENTS

4.1. Statistical analysis

Results will be presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The chosen dependent variable was the term q featuring each walking/running trial. The independent variables were age group (y), progression speed (m/s) and gradient (%).

Effects of gender and age on the dependent variable q were assessed by using a one-way ANOVA for unrelated measures. In addition, a post-*hoc* Bonferroni test was used to detect the strength of the associations between the dependent variable and gender/age. Moreover, effects of speed and gradient were assessed by using a one-way ANOVA for related measures. In addition, a post-*hoc* paired *t*-test (with Bonferroni correction) was used to detect differences between the dependent variable and speed/gradient. SPSS software (version 12.0 for Windows) was used for statistical analysis (Zakeri et al., 2006; Houdijk et al., 2009).

The graph legend is the same as already illustrated and described in chapter 8 (par. 3).

The average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 (see also chapter 6, par. 2.1) have been considered. The average q values are contained in the enclosed CD (First Study, Chapter 12, Term compound q).

4.2. The term *q* as a function of age

If the term q is represented as a function of age (Figure 12.1), our results show that:

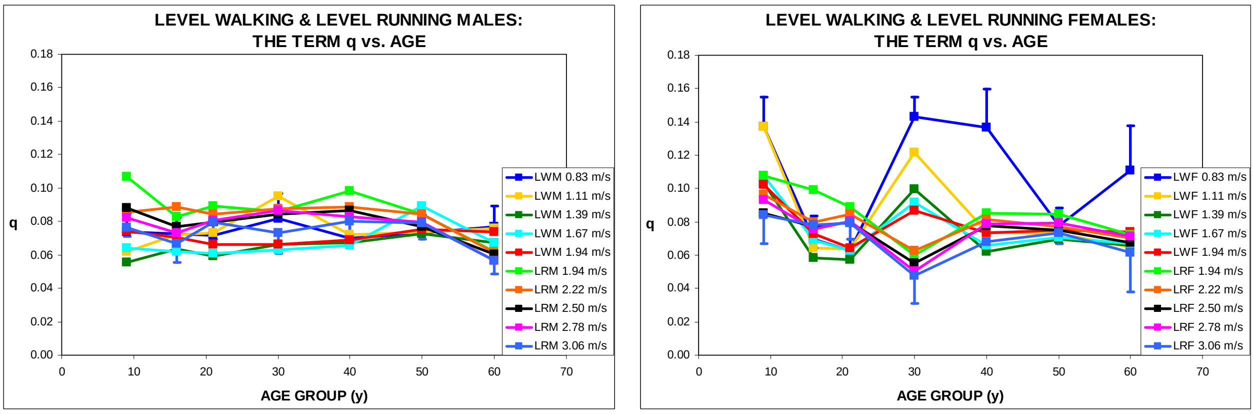

Figure 12.1. The term q as a function of age in level walking and running, males (on the left) and females (on the right).

- in males (left graph), both in level walking and running, *q* is not dependent on age. Indeed, its pattern is very regular and constant;
- in females (right graph), both in level walking and running, *q* is not so dependent on age. Yet, qualitatively, its pattern seems to be not so regular and constant: this is particularly evident in females aged 25 to 35 and in level walking at the lowest speed of 0.83 m/s.

Specific results of the statistical analysis (with relevance) are shown in Table 12.3.

SUBJECTS	**LEVEL WALKING AND RUNNING at each speed**
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 12.3. The term *q* as a function of age in level walking and running (males and females).

4.3. The term *q* as a function of speed

Moreover, if the term *q* is represented as a function of speed (Figure 12.2), our results show that:

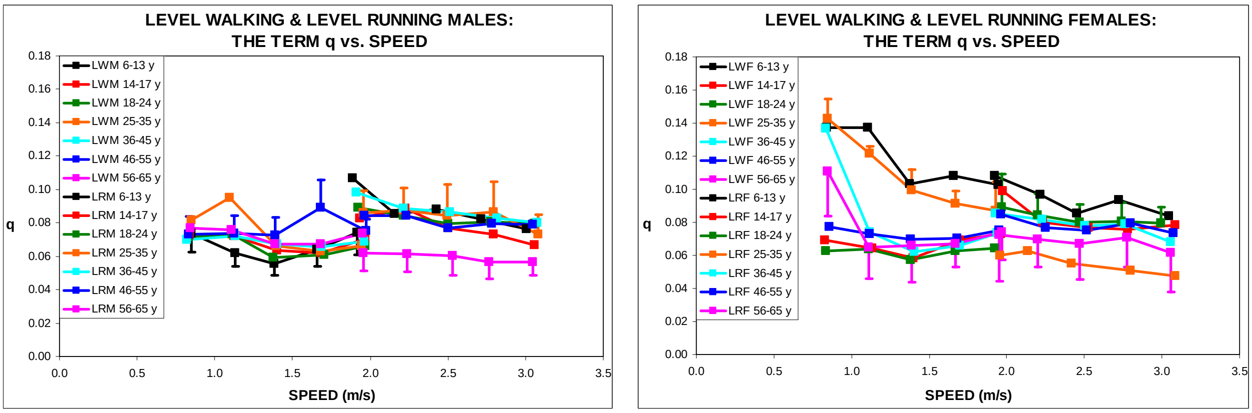

Figure 12.2. The term *q* as a function of speed in level walking and running, males (on the left) and females (on the right).

- both in walking and running, the term *q* does not significantly change with speed (and stride frequency; Cavagna et al., 1986; 1988; 1991; 1997; Minetti, 1998; Schepens et al., 1998; Cavagna et al., 2008), regardless of gender and age group;
- as shown, this pattern occurs both in males (left graph) and females (right graph);
- in detail, young females (aged 6 to 13) seem to have the highest values of *q*. This is probably coupled with a higher extension of the lower limbs. Furthermore, in running, *q* reaches its lowest value in males aged 56 to 65 and in females both aged 25 to 35 and 56 to 65. Conversely, this is probably coupled with a lower extension of the lower limbs.

Specific results of the statistical analysis (with relevance) are shown in Table 12.4.

SUBJECTS	LEVEL WALKING AND RUNNING in each age group
Males	p=NS comparing all speeds
Females	p=NS comparing all speeds

Table 12.4. The term *q* as a function of speed in level walking and running (males and females).

4.4. The term *q* as a function of gradient

Finally, if the term *q* is represented as a function of gradient (Figure 12.3), our results show that:

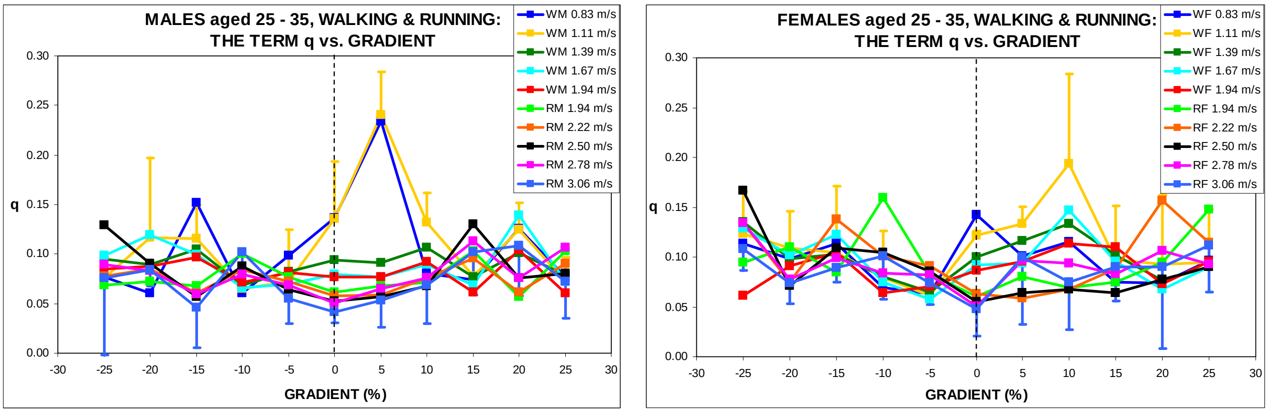

Figure 12.3. The term *q* as a function of gradient in walking and running, males (on the left) and females (on the right).

- the term *q* does not significantly change with gradient both in walking and running conditions;
- qualitatively, in males (left graph) it seems to be no significant differences as a function of gait speed, but in females (right graph) *q* seems to decrease as a function of walking/running speed.

Specific results of the statistical analysis (with relevance) are shown in Table 12.5.

SUBJECTS	GRADIENT WALKING AND RUNNING at each speed
Males	p=NS comparing all gradients
Females	p=NS comparing all gradients

Table 12.5. The term *q* as a function of gradient in walking and running (males and females).

4.5. Discussion

Previous to the present model, an important attempt to evaluate *q* was made by Minetti (1998), both in humans and horses locomotion. As stated in Equation [12.2], *q* values reflect the inertial properties of limbs. Particularly, the lower *q* represents a way to reduce the internal work by bending the limbs; thus, a consistent decrease of *q* at high speeds could be interpreted as an energy minimization strategy. Yet, this capacity in reducing the term *q* has been demonstrated only in horse locomotion. Indeed, horses display a much greater decrease when passing from walk to trot and then to gallop (Minetti, 1998).

In humans and horses, Minetti showed how *q* values resulted to be almost constant (= 0.100 ± 0.013) throughout all speeds (left graph), gradients (middle graph) and gaits (right graph), as expected (Figure 12.4). Indeed, the geometry in the middle of the stance and swing phases is similar as gaits, speeds and gradients change.

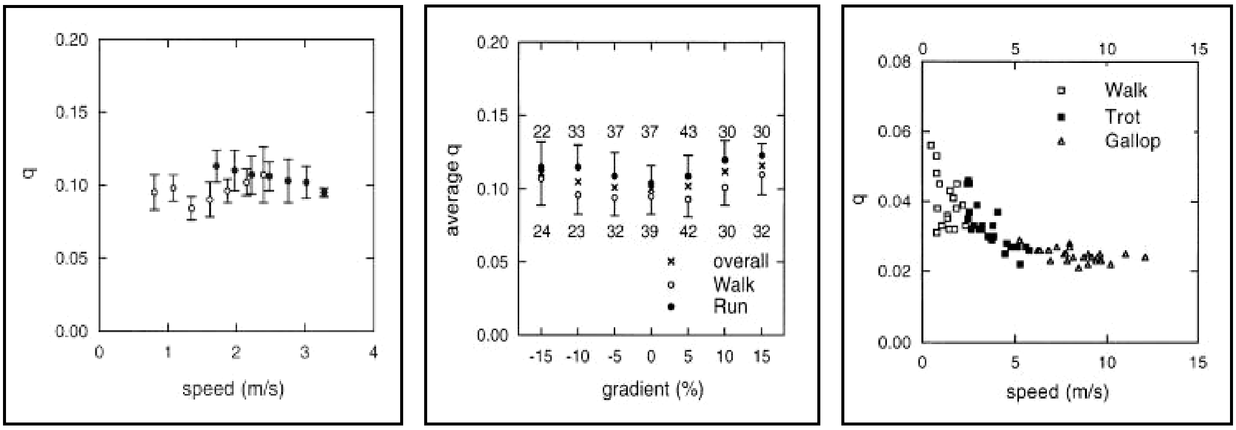

Figure 12.4. Average *q* in humans (on the left and in the middle) and horses (on the right) locomotion, in Minetti (1998).

In our research, we have tried to verify and demonstrate the validity and the applicability of the constancy of *q* in different age groups (from 6 to 65 years), in males/females who walked/ran at different speeds, both at the level gradient and/or at extreme slopes (downhill and uphill).

The constancy of the term *q* proved by Minetti (1998) has been only partially confirmed over our data. Particularly:

a) in males, the average *q* value at the level gradient is 0.070 ± 0.012, in walking; and 0.082 ± 0.014, in running, regardless of age and speed;

b) similarly, in females, the average *q* value is 0.085 ± 0.013, in level walking; and 0.079 ± 0.015, in level running;

c) therefore, independently of gaits, the final average value of *q* is 0.076 ± 0.013, in males; and 0.082 ± 0.014, in females.

Consequently, it makes clear that the difference between males and females is not so statistically significant. Moreover, despite the consistently negative trend of stride frequency (see also chapter 10, par. 2) and the positive trend of duty factor (see also chapter 10, par. 5) with respect of speed in young children (aged 6 to 13) and in elderly adults (aged 56 to 65), *q* values result to be constant regardless of age and gait, according to Minetti (1998).

To sum up, in level gaits, the average value of *q* seems to be about 0.08. Thus, the reference equation to estimate internal work (PW_{int}) of level human locomotion (according to Equation [12.1], described in par. 3.1 above) could be written as:

$$PW_{int} = SF \bullet s \bullet (1 + (\frac{DF}{1 - DF})^2) \bullet 0.08 \text{ [Eq. 12.4a]}$$

The discrepancy of our term *q* (with respect to Minetti's value) could be probably due to our lower values in Measured mechanical internal work. Furthermore, especially in females, our results strongly confirm the statement that 'the slightly decrease in *q* during running could reflect a greater knee flexion at high velocities, thus a reduced angular momentum during the swing phase' (Minetti, 1998).

Minetti (1998) stated that a similar value of *q* was found if the same analysis was extended to gradients ±15% (step 5%). The test protocol we adopted extended and completed the range of gradients investigated (±25%, step 5%).

Particularly, it seems that our results wholly confirm this statement. In fact:

d) in males, the average *q* value is 0.092 ± 0.019, in gradient walking; and 0.095 ± 0.017, in gradient running;

e) in females, the average *q* value is 0.125 ± 0.018, in gradient walking; and 0.066 ± 0.014, in gradient running, regardless of gradient and speed. In walking, the higher values of *q* could be due to a higher fat mass. Differently, in running, the smaller values could be related to a greater vertical displacement of the body centre of mass combined with a greater knee flexion (Minetti, 1998);

f) therefore, independently of gaits, the final average value of *q* is 0.093 ± 0.018, in males; and 0.095 ± 0.016, in females.

Therefore, the reference equation to estimate internal work (PW_{int}) of gradient human locomotion (according to Equation [12.1]) could be written as:

$$PW_{int} = SF \bullet s \bullet (1 + (\frac{DF}{1 - DF})^2) \bullet 0.10 \text{ [Eq. 12.4b]}$$

As hypothesized, in gradient gaits, our values of *q* wholly satisfy Minetti measurements (1998).

5. PREDICTED MECHANICAL INTERNAL WORK

5.1. Mathematical calculation

The main simple biomechanical variables (stride frequency and duty factor) and the progression speed were obtained by the aforementioned custom-written software in LabVIEW (see also chapter 6, par. 2.1). The term *q* has been calculated as previously described (see par. 4.5 above and relative mathematical equations). As a consequence, all these measurements were then converted into the mathematical model equation to calculate the Predicted mechanical internal work (see Equation [12.4]).

5.2. Statistical analysis

Results will be presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

Effects of gender, age, speed and gradient of the dependent variable PW_{int} have been calculated applying the same statistical analysis already described in par. 4.1 above.

Furthermore, effect of PW_{int} on the dependent variable MW_{int} was assessed by using a one-way ANOVA for related measures with an additional post-*hoc* Tukey test (with Bonferroni correction).

The graph legend is the same as already illustrated and described in chapter 8 (par. 3).

5.3. Results of our experiments

5.3.1. Predicted internal work as a function of age

If PW_{int} is represented as a function of age (Figure 12.5), our results show that:

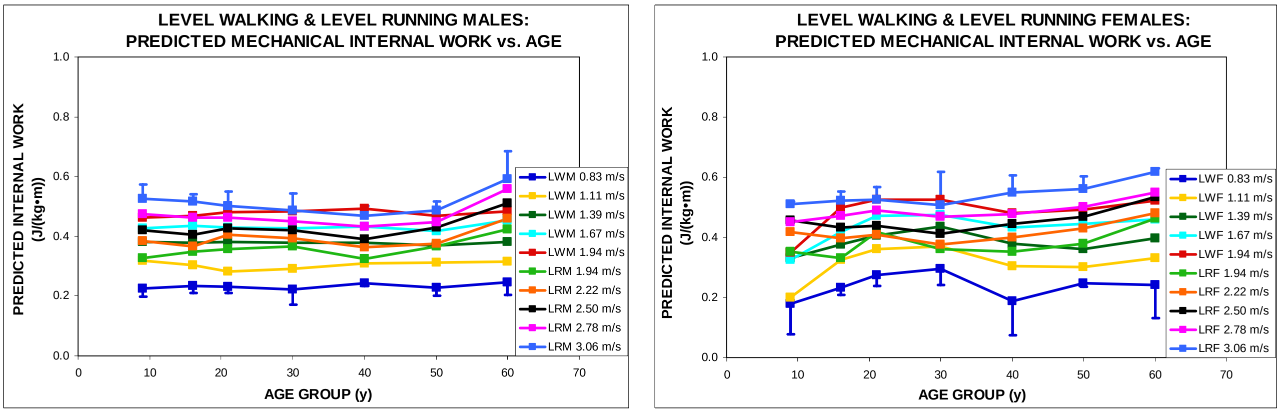

Figure 12.5. PW_{int} as a function of age in level walking and running, males (on the left) and females (on the right).

- as illustrated and demonstrated for Measured W_{int} (chapter 10, par. 8.2.1), both in level walking and running, PW_{int} is not dependent on age;
- this pattern occurs both in males (left graph) and females (right graph);
- finally, at the most walking speeds, PW_{int} is always lower than PW_{int} in running.

Specific results of the statistical analysis (with relevance) are shown in Table 12.6.

SUBJECTS	**LEVEL WALKING AND RUNNING at each speed**
Males	p=NS comparing all age groups
Females	p=NS comparing all age groups

Table 12.6. PW_{int} as a function of age in level walking and running (males and females).

5.3.2. Predicted internal work as a function of speed

The average value q (= 0.08) for level walking/running has been computed and fed back into Equation [12.1] to predict PW_{int} from individual data on speed, stride frequency and duty factor.

If PW_{int} is represented as a function of speed (Figure 12.6), our results show that:

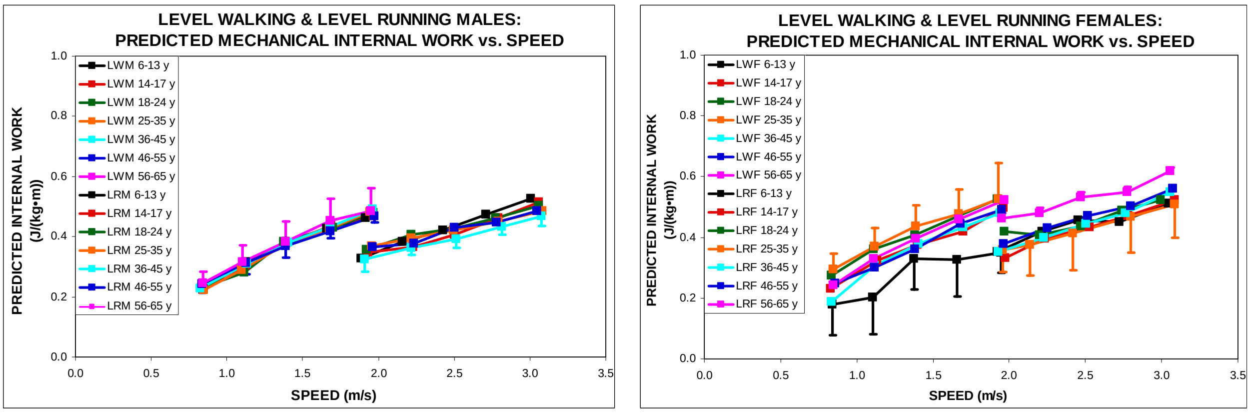

Figure 12.6. PW_{int} as a function of speed in level walking and running, males (on the left) and females (on the right).

- as previously illustrated and demonstrated for Measured W_{int}, PW_{int} is highly dependent on speed: if walking or running speed increases, it subsequently increases ($p<0.001$);
- this pattern occurs similarly in each testing condition independently of gender, age, type of locomotion or speed;
- in addition: a) subjects (males and females) aged 6 to 13 have the lowest value of PW_{int} due to their highest value of the term q (in running); conversely b) subjects (males and females) aged 56 to 65 have the highest value of PW_{int} due to their lowest value of the term q.
- all our results concur with literature data (Minetti, 1998: Figure 12.7).

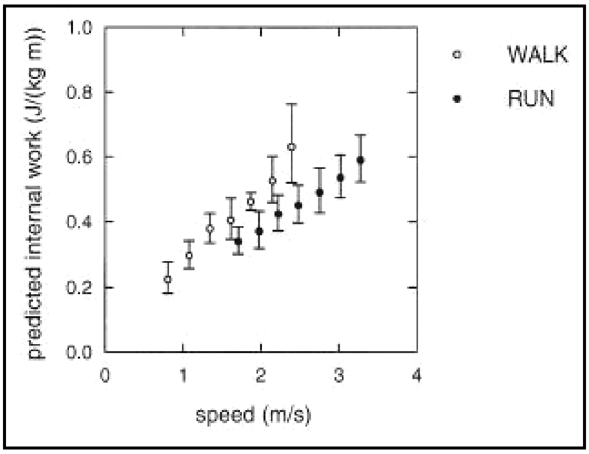

Figure 12.7. PW_{int} as a function of speed, in Minetti (1998).

Specific results of the statistical analysis (with relevance) are shown in Table 12.7.

SUBJECTS	**LEVEL WALKING AND RUNNING in each age group**
Males	$p<0.001$ comparing all speeds
Females	$p<0.001$ comparing all speeds

Table 12.7. PW_{int} as a function of speed in level walking and running (males and females).

5.3.3. Predicted internal work as a function of gradient

The average value *q* (= 0.10) for gradient walking/running has been computed and fed back into Equation [12.1] to predict PW_{int} from individual data on speed, stride frequency and duty factor.

If PW_{int} is represented as a function of gradient (Figure 12.8 and 12.9), our results show that:

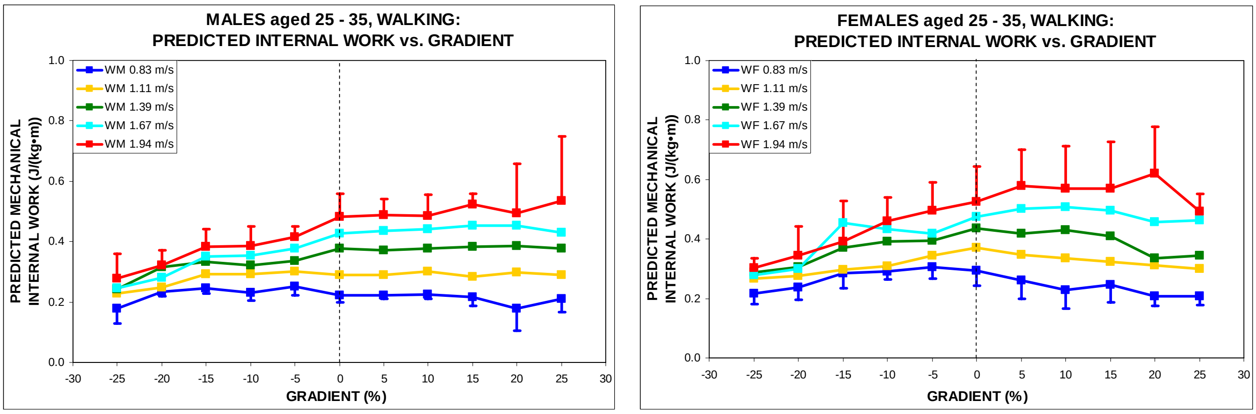

Figure 12.8. PW_{int} as a function of gradient in walking, males (on the left) and females (on the right).

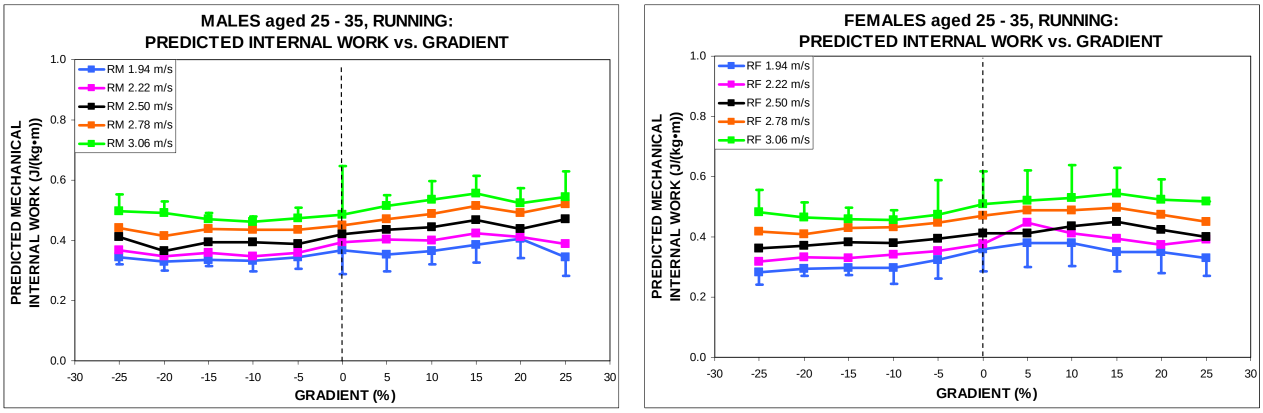

Figure 12.9. PW_{int} as a function of gradient in running, males (on the left) and females (on the right).

- in walking, PW_{int} is only slightly dependent on gradient ($p<0.05$). This pattern occurs similarly in each testing condition independently of gender, type of locomotion or speed;
- differently, in running, PW_{int} is not dependent on gradient;
- moreover, this pattern occurs similarly both in males (left graph) and females (right graph).

Specific results of this statistical analysis (with relevance) are shown in Table 12.8.

SUBJECTS	GRADIENT WALKING at each speed	GRADIENT RUNNING at each speed
Males	p<0.05 comparing all gradients	p=NS comparing all gradients
Females	p<0.05 comparing all gradients	p=NS comparing all gradients

Table 12.8. PW_{int} as a function of gradient in walking and running (males and females).

To sum up, we can conclude that, as well Measured internal work (MW_{int}), PW_{int} is more dependent on speed than on age and gradient.

Its average values are contained in the enclosed CD (First Study, Chapter 12, Template Predicted W_{int} and average PW_{int}, in level and gradient gaits).

6. PREDICTED INTERNAL WORK *VERSUS* MEASURED INTERNAL WORK

6.1. Level walking and running

Figure 12.10 shows PW_{int} pattern as a function of MW_{int} both in level walking and running, in males (left graph) and females (right graph).

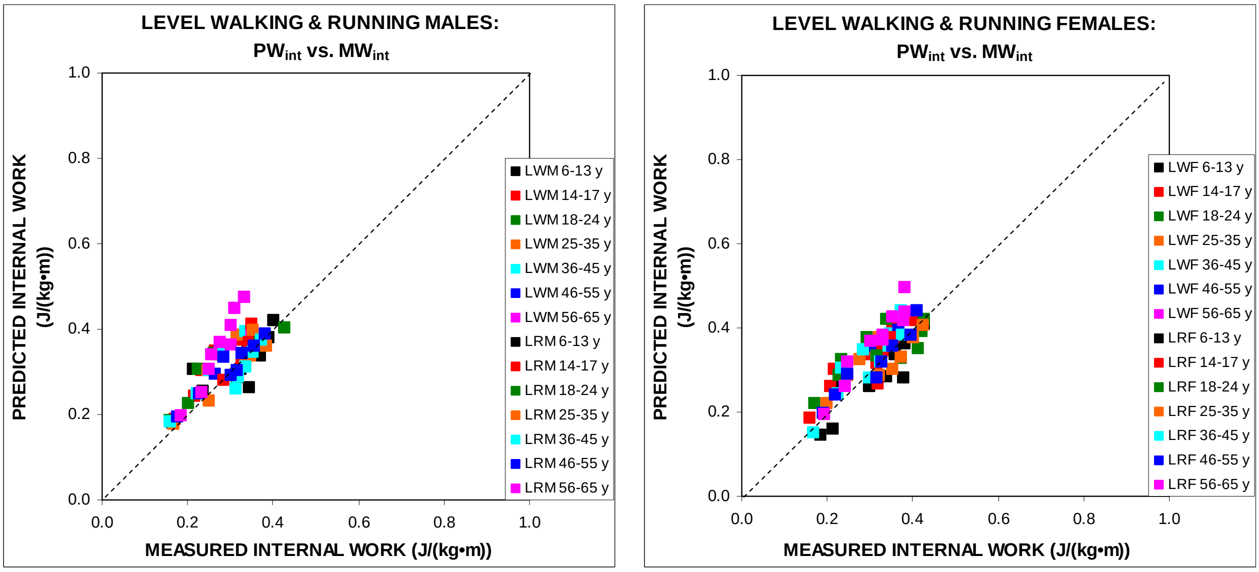

Figure 12.10. PW_{int} *versus* MW_{int} in level walking and running, males (on the left) and females (on the right).

Both in males (n = 70, R^2 = 0.617, r = 0.785) and females (n = 70, R^2 = 0.620, r = 0.787), average values of PW_{int} are very close to the corresponding MW_{int}. This pattern is similar in walking and running gaits.

Yet, our results underestimates the predicted internal work if compared to Minetti's data (Figure 12.11). Importantly, some possible explanations of the distance of our internal works in comparison to Minetti's values (1998) could be found in: a) a different sampling frequency (100 Hz *versus* 120 Hz); and b) the different filtering procedures we have applied in the kinematic data analysis (i.e. a 'non adaptive filtering' instead of a 'local adaptive filter' based upon the mathematical algorithm described in Ferrigno et al. (1990) and Borghese et al. (1991)).

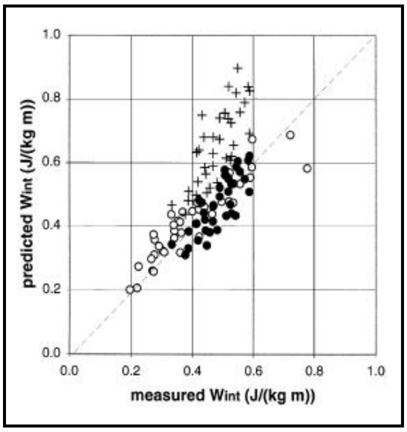

Figure 12.11. PW_{int} *versus* MW_{int} in level walking and running, in Minetti (1998).

6.2. Gradient walking and running

In the graphs below, blue represents the relationship between the independent and the dependent variable during walking and running at -25%; grey at -20%; yellow at -15%; dark green at -10%;

pink at -5%; brown in the level condition; sky-blue at +5%; orange at +10%; red at +15%; green at +20%; and black at +25%.

Figure 12.12 shows PW_{int} pattern as a function of MW_{int} both in gradient walking and running, in males (left graph) and females (right graph).

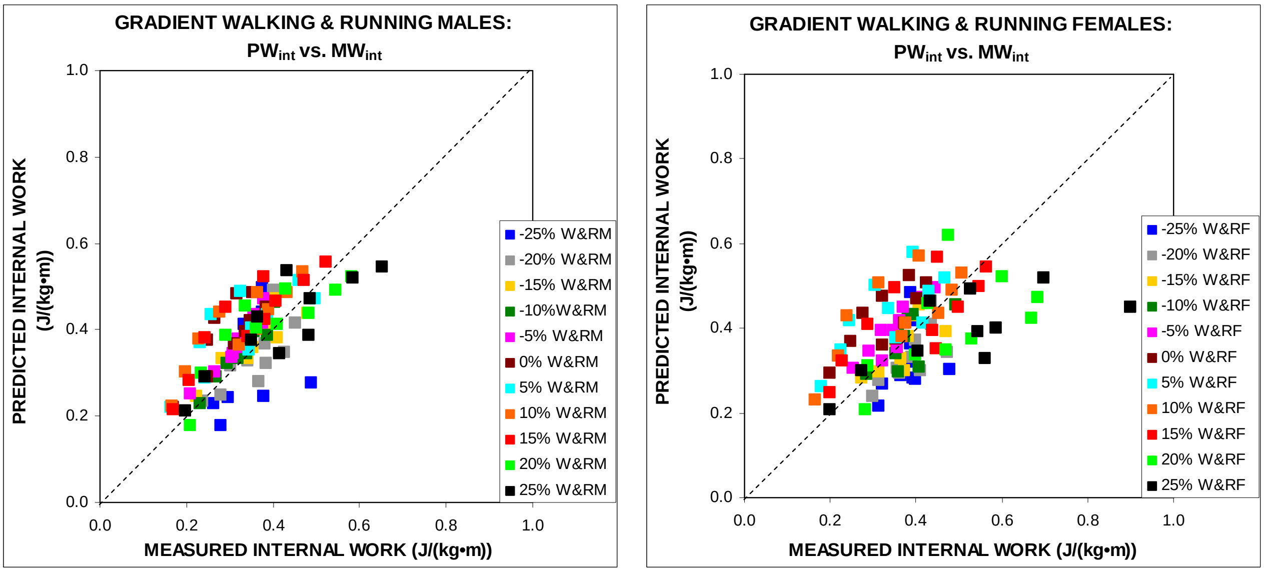

Figure 12.12. PW_{int} *versus* MW_{int} in gradient walking and running, males (on the left) and females (on the right).

Both in males ($n = 110$, $R^2 = 0.525$, $r = 0.724$) and females ($n = 110$, $R^2 = 0.295$, $r = 0.543$), average values of PW_{int} are close to the corresponding MW_{int} values, as well. This pattern is similar in gradient walking and running gaits. Yet, our results underestimates the predicted internal work if compared to Minetti's data (i.e. the different filtering procedures applied in the kinematic data analysis).

6.3. Discussion

As expected, in level gaits, the q values are quite constant at different speeds (for both genders).

Among the others, this implies that the Predicted internal work calculated by assuming a constant q (= 0.08) well concurs with the experimentally Measured work. Therefore, the match between the two methods is very close.

In gradient gaits, the q values are quite constant at the different speeds only in males. Yet, observing the results, we have decided to assume a constant q (= 0.10) both in males and females. Consequently, the Predicted internal work well concurs with the experimentally Measured work. Therefore, the match between the two methods is close (excepted few points), too.

In conclusion, both at the level and downhill/uphill gradients, the direct comparison between Predicted and Measured internal work has shown the close relationship among the two.

Moreover, we have tried to compare our Predicted values (obtained by applying Minetti's model equation, 1998) to values obtained by applying another mathematical model equation. Specifically, the equation we used has been proposed in Cavagna et al. (1991) and recently discussed in Cavagna et al. (2008). This model has been validated only in running gaits:

$$PW_{int_C} = 0.1440 \bullet 10^{-(2 \bullet SL)} \bullet SF \bullet s \quad [Eq.\ 12.5]$$

where PW_{intC} represents the Predicted mechanical internal work (J/(kg·m)); *SL* the stride length (m); *SF* the stride frequency (Hz); and *s* the average speed progression (m/s).

In such a way, it will be possible to compare two different model equations stating their similarities/differences. By comparing these two work equations, we could observe that:

- Cavagna's empirical equation (1991) takes into account only 3 variables: two of them are independent (the stride frequency, SF, and the progression speed, s), and the last one has been obtained from the previous ones (the step length, SL);
- Minetti's predictive equation (1998) represents a more complete model in which each single variable has its own independence: the stride frequency (SF), the progression speed (s), the duty factor (DF) and the compound dimensionless term (*q*). This model also considers the real pattern of movement.

Consequently, we think that both the validation and the application of this last equation could fit well the experimental data obtained by cinematographic recordings.

The graph legend is the same as already illustrated and described in chapter 8, par. 3. Differently, in this case, W_{int} ranges from 0 to 0.6 J/(kg·m) (step 0.2 J/(kg·m)).

A. In level running gait, our results show that, independently of both age and speed, Minetti's equation overestimates predicted internal works in comparison to Cavagna's model (Figure 12.13a). Indeed, this discrepancy reaches 38.5% in males and 37.5% in females. Clearly, this overestimation increases with walking/running speed. Furthermore, a wider vertical scattering has been observed in works derived by applying the Minetti's method.

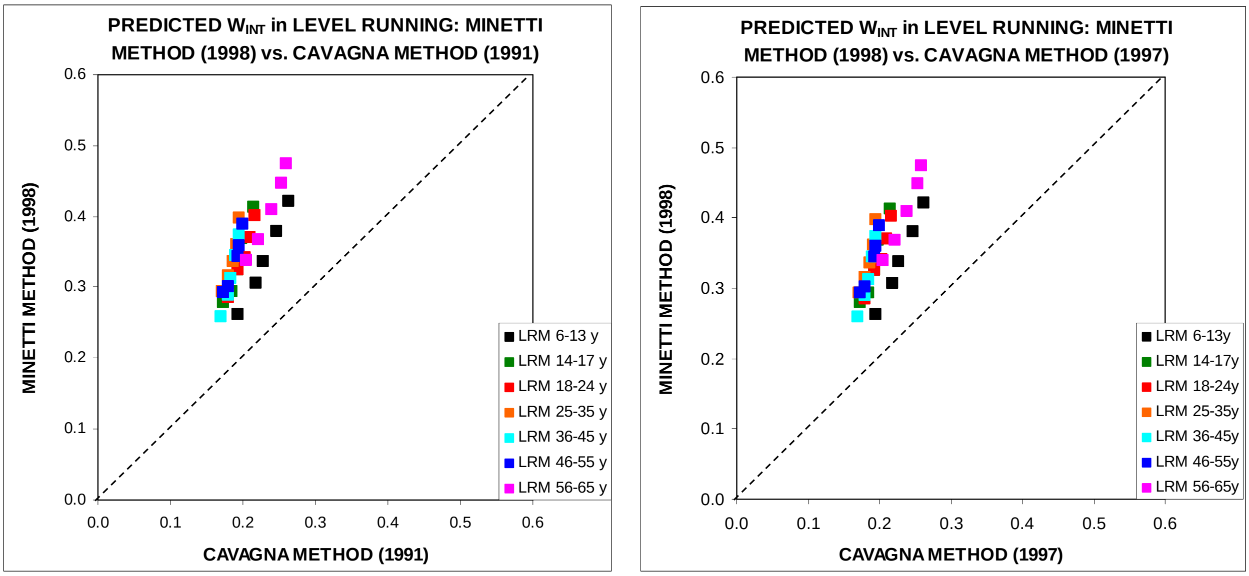

Figure 12.13a. PW_{int} in Minetti's method (1998) *versus* PW_{int} in Cavagna's method (2008) in level running, males (on the left) and females (on the right).

B. In gradient running gait, our results confirm the previous statement showing that, independently of both gradient and speed, Minetti's equation overestimates predicted internal works in comparison to Cavagna's model (40.5% in males and 38.4% in females: Figure 12.13b). As at the level condition, a wider vertical scattering has been observed in works derived by applying the Minetti's method.

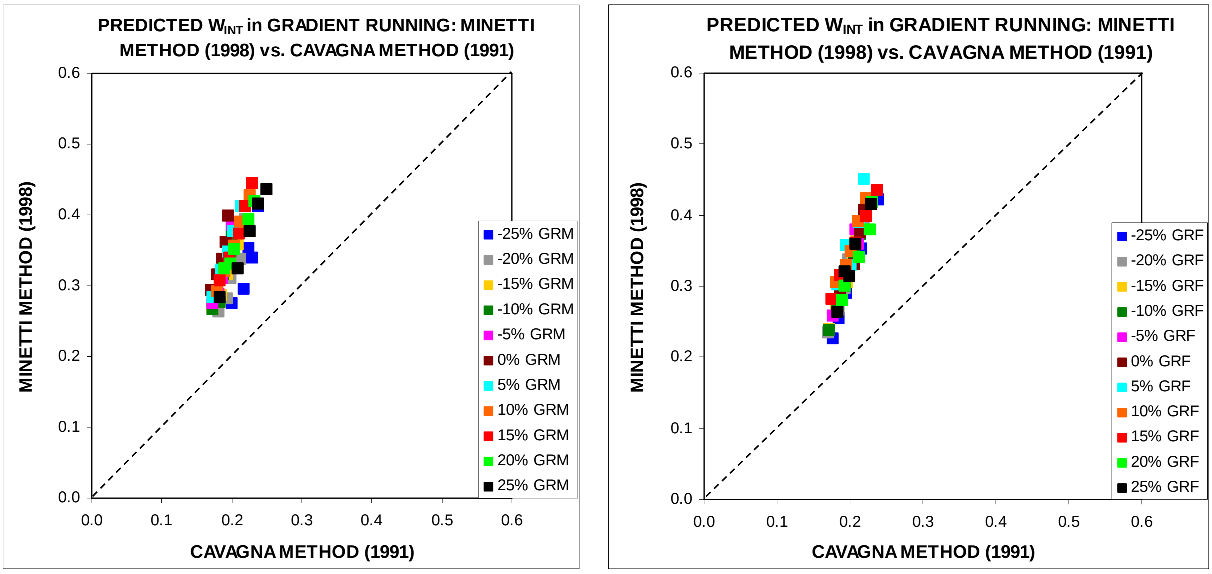

Figure 12.13b. PW_{int} in Minetti's method (1998) *versus* PW_{int} in Cavagna's method (2008) in gradient running, males (on the left) and females (on the right).

Theoretically, a possible explanation of the distance of our internal works in comparison to Cavagna's values (1991) could be found: **a)** in the terms involved in the mathematical equation; **b)** in the absence of a filtering procedure applied upon raw kinematic data; **c)** different filtering procedures that have been applied in the kinematic data analysis ('non adaptive filter', 5th order Butterworth filter with a 8.5 Hz cut off frequency, instead of 'local adaptive filter'; for more details see also Savitzky et al., 1964); **d)** the progression speed has been inferred from the foot contact period on the treadmill; **e)** according to Minetti et al. (1993), the positions and movements of the body centre of mass have been measured relative to the body segments; **f)** the angular velocity of each segment was calculated from the displacement curves, giving, as a function of time, the angle made by the segment with the horizontal; **g)** therefore, the kinetic energy of each segment, due to their velocities relative to the body centre of mass (and not to shoulder and hip joints, as in Fenn, 1930, Cavagna et al., 1977; 1991) has been calculated as the sum of translational and rotational energy. Then, the internal power has been obtained as the sum of the increments of the total energy during a single stride; and **h)** as a result, the mechanical internal power has been calculated separately in each limb by respecting the 'intra limb exchange' (according to Minetti et al., 1993; 1994) (see Appendix).

7. CONCLUSION

Results of this study have demonstrated that the direct measurement of mechanical internal work can be compared to the indirect measurement, in human locomotion. On one hand, it is important to remember that different values of the term q have to be used at level or gradient gaits. On the other hand, it appears that the two methods (PW_{int} *versus* MW_{int}) give very similar results.

Importantly, the scientific mathematical prediction of the mechanical internal work provides accurate estimations as far as gender, age, gait, speed and gradient are concerned in human locomotion. By having been applied for a wide number of conditions, the model equation (together with the corresponding q values) seems suitable to be appropriately used whenever the direct experimental measurement is not available.

Areas for further research may include similar analysis for other age groups, like younger subjects (< 6 years) for their different locomotion techniques and lower limb length, and older ones (> 65 years) for their physiological constraints due to the increased age.

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APPENDIX

This Appendix briefly summarizes two important works (Minetti et al., 1992; Minetti, 1998) which play a key role in better understanding the origin of both Equations [12.1] and [12.2] importantly involved in our model.

Minetti et al. (1992) schematized a stiff limb diagram model adopted to formalize the mechanical internal work model. The main assumptions within this model are: stiff limbs, no double support, body centre of mass located in the hip joint. Precisely, four stiff segments (two lower and two upper limbs) are involved in the computation. The lower limb length, the proximal distance of the lower limb centre of mass, the upper limb length (as a fraction of the lower limb length), the radii of gyration of lower and upper limbs have been taken from the literature. Assuming that the limb extremities follow a sinusoidal displacement with respect to the body centre of mass, which is placed in the head-trunk segment and does not move horizontally because of the symmetrical positions of the limbs, the mechanical internal work (W_{int}) has been evaluated from the oscillations in kinetic energy (both translational and rotational) according to the König theorem, as suggested in Cavagna and Kaneko (1977). Indeed, the energy transfer among segments is not relevant because of the in-phase shapes of the kinetic energy curves.

Therefore, Minetti and Saibene (1992) have modelled a mathematical equation to calculate the work needed to accelerate the body limbs with respect to the body centre of mass as:

$$W_{int} = SF \bullet s^2 \bullet \frac{\pi^2}{2} \bullet [(a^2 + g^2)(m_L + b^2 m_U)] \quad \text{[Eq. A1]}$$

where *SF* is the stride frequency (Hz); *s* is the average progression velocity (m/s); *a* represents the average proximal distance and *g* the mean radius of gyration of the limb centre of mass (as a fraction of limb length); m_L and m_U refer to the lower and upper limb with the respect to the body mass (kg); and *b* is the upper limb length (as a fraction of the lower one).

In the Equation [A1] equal amounts of time spent by the limbs in the stance and the swing phase have been considered. Clearly, such assumption is not available when different gaits and speeds have to be simultaneously taken into account. Furthermore, it has been supposed that lower and upper limbs show similar radii of gyration.

Experimentally, in W_{int} measurements, the time course of the limb kinetic energy shows two different peaks, one related to the progression speed (s_{ST}, the top speed during the stance time), and the other reflecting the limb speed, relative to the overall centre of mass, during the swing phase

(s_{SW}). Therefore, the reference to speed in the equation above can be modified to accommodate this rationale by replacing

$$s^2 = \frac{1}{2}s^2{}_{ST} + \frac{1}{2}s^2{}_{SW} \text{ [Eq. A2a]}$$

and by considering that s_{ST} is equal to s, and s_{SW} can be calculated by introducing the 'duty factor' as

$$s_{SW} = s_{ST} \bullet (\frac{DF}{1 - DF}) \text{ [Eq. A2b]}$$

This means that the limb speed, with respect to the centre of mass, during the swing phase is higher than the one during the stance when *DF* is greater than 0.5, as in walking. By including Equations [A2b] and [A2a] into Equation [A1], we obtain the initial Equation [12.1]:

$$PW_{int} = SF \bullet s \bullet (1 + (\frac{DF}{1 - DF})^2) \bullet q \text{ [Eq. 12.1]}$$

For the single terms' significance see par. 3 above.

PART 3

CONCLUSIONS

Chapter 13

CONCLUSIONS

1. INTRODUCTION

The repeatability of gait variables measures (Cappozzo et al., 1984) is of great relevance in clinical/sport fields. Indeed, this will develop and characterize both qualitative and quantitative gait analysis's 'multilateral' approach (Kadaba et al., 1989). Importantly, it is crucial to ask whether or not results from a single gait evaluation is representative of a subject's (or population's) overall gait performance and whether the data are consistent enough from day to day for taking significant clinical/training decisions.

On one side, clinical gait analysis has become widely utilized in the assessment of pathological gait: for instance, children and adolescents with cerebral palsy, amputation, degenerative joint disease, poliomyelitis, multiple sclerosis, dystrophy, myelomeningocele, rheumatoid arthritis and brain traumas whose movements are often complex, multi-planar and distorted (Davis, 1997). Furthermore, clinical gait laboratories often compare the walking characteristics of individuals with orthopaedic or neurological pathologies to the walking patterns of healthy individuals (Zajac et al., 2003).

On the other side, the experienced eye' keeps the most prominent role in prognostic/diagnostic process in sport context, as well (Cavagna et al., 1981; Slawinski et al., 2004).

As a consequence, the comparison of the walking kinematics, kinetics and muscle activity patterns between healthy and impaired or improved individuals serves as the basis for defining abnormal gaits. Therefore, based upon previous literature (Minetti, 2006a; 2006b), we have investigated the locomotor pattern (i.e. the three-dimensional trajectory of the relevant but often neglected gait analysis variable 'centre of mass of the human body') in healthy subjects (males and females) moving in various gaits, speeds and gradients.

The main novelty of our study is the development (and the application) of a mathematical method (Fourier Series) which could better describe (and graphically represent) each peculiar individual (subject or population) gait signature. In this way, it becomes easy to assess, in each movement direction, the right/left symmetry in BCOM trajectory between the two stride phases. The knowledge of both simple and complex biomechanical variables fully completes the planning of our research project.

In the next sections, we will briefly discuss its main peculiarities, limits and future developments. However, in depth, at the end of each chapter single results have been widely discussed and commented according to previous studies and references.

2. MAIN PECULIARITIES

New gait analysis protocols (and models) have been introduced as evaluation tools in health, sport and clinical context (Sutherland, 2001; 2002; Slawinski et al., 2004; Sutherland, 2005).

All our measurements (i.e. recordings of kinematic data) have been made by the same subject thus avoiding operator-dependent errors. Furthermore, the same climatic conditions (i.e. humidity and temperature) have been kept.

In our research project, a wide number of moving conditions has been tested: males and females of different ages who walk and run at different speeds both at level and gradient slopes. Indeed, we want to verify our initial hypothesis that at every gender, age, gait, speed and gradient, a unique 3D *contour* is associated (Minetti, 2006a). Positively, this aim has been wholly confirmed.

The test protocol we administered has been successfully carried out and completed by the majority of subjects. Indeed, we had to reject just a limited number of trials.

Specifically, we have established 1) a mathematical continuous method (or function) and 2) a valid evaluation protocol for the study of 3D BCOM pattern during human locomotion over time and space (Hildebrand, 1967). This mathematical method (Fourier Series) and its graphical representation (Lissajous *contours* and polar-logarithm graphs) permitted us to characterize:

1. the specific individual Digital Locomotory Signature (DLS);
2. the right/left (a)symmetries (SI) (Shorter et al., 2008).

Importantly, we built up an initial comprehensive *database* of ‘normality values’ (reference equations/coefficients) describing normal locomotion in a set of different conditions. Clearly, the knowledge and the development of this approach constitute the most important and relevant theoretical and practical results of this study. Indeed, it might be possible to extend the main advantages of such protocol, data analysis and results to both impaired and improved gaits.

Furthermore, our results have shown that:

a) right and left sides of the stride are quite asymmetrical. Globally, this asymmetry is probably related both to anatomy (i.e. leg length) and which hand the studied people use (i.e. predominant right handedness). Indeed, the precise characterization of such physiological limitations of the (a)symmetry in healthy people is an important plus of our study;

b) in each testing condition, males and females are quite similar;

c) as expected, on average, the symmetry pattern is slightly lower in running gaits;

d) as expected, young children and elderly adults are the most asymmetrical subjects, independently of testing conditions. This major asymmetry could be due to the development of gait process in children (Gurney, 2002) and to structural wearing down of musculoskeletal system in old adults (Magnusson et al., 2008).

In addition, the implementation of programs in LabVIEW software to deeply analyse kinematic data constitutes an important step for further researchers who want to develop and apply the same approaches. Specifically, the Lissajous graphical *contours* visualize well how the average 3D displacement of the BCOM moves, veers and turns in all planes. The graphical arrangement of two (or more) *contours* makes the individuation (and classification) of their specific peculiarities and differences easier. Moreover, the graphical simultaneous combination of both amplitude and phase coefficients in a polar logarithm graph facilitates definition and characterization as a function of gait, speed and gradient.

Finally, the investigation of both simple and complex biomechanical variables in such a large number of moving conditions has wholly confirmed and extended the previous documented data (Sibley et al., 2009). Moreover, both the application of the mathematical method to calculate biomechanical variables (by using potential, kinetic and total energies) and the revision of the mathematical method to calculate internal work (model equation in Minetti, 1998) represent open sources for further researches.

3. MAIN LIMITATIONS (OR DISADVANTAGES)

Despite the great number of testing conditions we have investigated, some important limitations in our study could find out.

The high similarity between males and females (both in BCOM *contour* and symmetry) could represent an initial restraint. Indeed, we suggested there might be significant differences related to gender (i.e. in the BCOM excursion). They were probably due to a different anatomy. Contrary to what was expected, our results differ to our initial hypotheses maybe because of the strong match in anthropometric parameters between males and females. So, further research will focus on younger children (< 6 years) and older adults (> 65 years): hopefully, the supposed gender differences will increase. Furthermore, children analysis might be fundamental in isolating and characterizing both toddler and skipping gaits. Indeed, these gaits seem to better illustrate locomotion in young subjects. In such a way, Froude number could be use as a distinctive biomechanical independent variable.

Positively, significant differences have been found as a function of age, gait, speed and gradient.

The test protocol we administered includes only a limited number of velocities (walking and running). To verify the greater right/left symmetry at increasing locomotion speed, a wider set of speeds has to be planned and administered. These further testes have to performed to confirm the optimal walking speed related to different ages, as well.

On the other side, gradient locomotion (walking and running) has been tested only in the middle-aged class (subjects aged 25 to 35). This will constitute a starting point for applying the same test protocol at different age groups according to their specific locomotion patterns (and correlated limitations). In this way, it will be easy to compare the effect of the factor 'gradient' in depth, defining the optimal walking speed during gradient gait, as well.

As for the equation coefficients, negatively, the statistical analysis has been applied only for the main age groups (children *versus* young adults *versus* elderly adults) regarding only the first four harmonics (e.g. amplitude coefficients and sine/cosine functions). Although a slight significance has been found, it will be crucial to investigate subjects' behaviour also in the other age groups. However, we suppose that the two last harmonic coefficients (5^{th} and 6^{th} ones) do not play an important role in determining the overall locomotor pattern. Consequently, they could be neglected in the statistical analysis.

As for the polar graphs, the range of phase standard deviations has been graphically represented only in the main age groups (males and females). To fully complete the graphical representation, the same graphs have to be drawn up for the other age groups, as well. Moreover, it could be necessary to mathematically define an overall standard deviation value, which regards all locomotion speeds.

As far as biomechanical variables have been concerned, only kinematic data has been recorded. However, it is known that both kinetics and electromyography are very important in determining the main steps (and functions) in gait analysis (Sutherland, 2001; 2005). So, in the near future, we will probably use a) an instrumented treadmill with force platforms on the inside (Racic et al., 2009) and b) electrodes to record the activity of the main muscles of the lower limbs and trunks (Zajac et al., 2003). Kinetics data could help in evaluating and calculating in different ways both external and internal (and total) mechanical work (Cavagna et al., 1977; Willems et al., 1995). These new data have to be compared both to results obtained by kinematics and literature.

Furthermore, external work and energy recovery percentage that have been calculated using Fourier coefficients have been investigated only at level conditions. An implementation and optimization of this mathematical analysis would simplify the characterization of these variables also at gradient gaits and along medial/lateral direction.

4. FUTURE DEVELOPMENTS

As already discussed, future perspectives include the ability to detect variations in locomotion dynamics such as those caused both by training (‘improved’ locomotion) and passive aids, ageing, gait pathology and rehabilitation (‘impaired’ locomotion; Reisman et al., 2007). Indeed, the quantitative evaluation of the changes in the global locomotion pattern, during and following training sessions and rehabilitation treatments, could help to better understand their overall effects and to modulate the interventions.

So, we would like to **a)** extend these experiments (with modifications to the test protocol according to subjects’ abilities) to other improved and pathological conditions, and **b)** spread the advantages of this method to detecting gait anomalies. Clearly, our results, reached by applying a mathematical method, could be compared to other results obtained:

1. with dynamic optimization, another powerful tool (Anderson et al., 2001);
2. by recording ground reaction forces and other kinetic parameters;
3. by improving simulations to identify the sources of pathological movement and establish a scientific basis for treatment planning (Delp et al., 2007).

To confirm the more global pronounced asymmetry in extreme ages, we will also investigate younger children (< 6 years) and older adults (> 65 years). Indeed, we presume that the youngest and the oldest subjects will present more asymmetrical and peculiar loops coupled with a lower right/left symmetry related to their different anatomy.

In addition, to extend our protocol and to confirm the statistical sample theory it might be necessary to involve a large number of subjects for each age group.

Moreover, to fully complete our initial *database* testing the key role of the handedness, we will study both healthy and pathological left-handed dominant subjects (belonging to different gender and age groups). Indeed, as witnessed by literature, we expect a different dominant side which will produce turning and veering in an opposite way and direction (Lund, 1930; Souman et al., 2009).

As previously discussed, the walk-run gait transition speed (and the optimal walking speed) will be investigated in depth in order to better understand gait developments related to increasing age (Archer et al., 2006).

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UNIVERSITY OF VERONA

Department of Neurological and Visual Sciences

PhD Program in Exercise and Human Movement Science

Cycle XXII°

SECOND STUDY

THE RELATIONSHIP BETWEEN BODILY SYMMETRIES AND RUNNING ECONOMY IN HUMANS

Chapter 14

OVERVIEW OF SECOND STUDY

The research literature has shown a significant relationship between static anatomical/kinematic functional symmetries and running economy only in animal models (e.g. horse face symmetry *versus* race result; Manning et al., 1991; 1994a; 1994b; Parkes et al., 2009).

Thus, we have tried to go on a step further and show this possible relationship in human models (Raibert, 1986; Manning et al., 1998; Karamanidis et al., 2003).

To analyse and verify various static anatomical and kinematic functional symmetries, we used MRI (Magnetic Resonance Imaging; see chapter 15) and the motion capture technique (see chapter 3), respectively. Running economy was recorded with the portable metabolic system K4b^2 (Cosmed; see chapter 15).

We were interested in male subjects of different ages (from 20 to 55 years), featuring different running abilities:

- occasional runners: training of 2 hours *per week* with marathon best time (MBT) > 5 hours;
- skilled runners: training from 2 to 6 hours *per week* with 3 hours < MBT < 5 hours;
- top runners: training up > 6 hours *per week* with MBT < 3 hours.

As a result, we created three test groups, based on this specific running ability (Bramble et al., 2004). All subjects volunteered for the study, were informed and gave their full consent prior to taking part to the tests. These participants had to have no impediments as far as neurological or musculoskeletal pathologies affecting running ability were concerned (for more details, see chapter 16). Level running was performed at different speeds (from 2.22 to 5.00 m/s, step 0.56 m/s).

All testing was carried out utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

As regards running economy, no significant differences among the three groups were found (for more details, see chapter 18).

As far as static anatomical symmetries were concerned, MRI scan were analysed by means of software *ad hoc* by applying both a two-dimensional (DicomWorks and ImageJ software) and three-dimensional (LabVIEW 8.6 software) approach (for more details, see chapter 17). However, only slight significant differences have been found among runners (for more details, see chapter 18).

As far as dynamical functional symmetries were concerned, each kinematic data has been elaborated by means of a custom-written LabVIEW software (Minetti et al., 1993). We have also graphically represented the so-called *Digital Locomotory Signature* (for more details, see chapter 6) and measured the so-called *Symmetry Index* (for more details, see chapter 8) in order to demonstrate that occasional runners are more asymmetrical than the habitual ones (skilled and top), especially in the forward movement direction, while in the other directions top runners are the most asymmetrical (Draper, 2000; for more details, see chapter 17 and 18).

In addition, a comparison between treadmill and over-ground running (occasional *versus* skilled runners) has been performed, as well (for more details, see chapter 19). Knowledge of these measures is important both to extract and characterize the individual gait signature and also to fully describe the mechanics of running.

Finally, both the biomechanics of running (simple and complex variables) and the step variability of the body centre of mass (in each movement direction) have been investigated (for more details, chapter 20).

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PART 1

MATERIALS AND METHODS

Chapter 15

INSTRUMENTATION

1. INTRODUCTION

In this chapter, we are going to focus on the two main different instruments we used in order to carry out all test experiments and protocols. As previously described in chapter 4, for each piece of equipment, we have submitted: a) a brief review from the literature available to define and evaluate the main characteristics (also in terms of advantages and disadvantages); and b) a brief and simple presentation and illustration of the specific components and relative functions.

The equipment is illustrated in this order:

1. Magnetic Nuclear Resonance (par. 2) and, of utmost importance, Magnetic Resonance Imaging (par. 3), to record static symmetries;
2. portable metabograph K4b^2 (par. 4), to record physiological parameters and running economy.

All trials were performed on the treadmill h/p/Cosmos (Saturn 4.0); for its characteristics and properties see also chapter 4, par. 2. Furthermore, Vicon motion capture system was used to record kinematics (dynamic symmetries); for its characteristics and properties see also chapter 4, par. 3.

2. NUCLEAR MAGNETIC RESONANCE

2.1. Generality

Nuclear Magnetic Resonance (NMR) is 'a physical phenomenon based upon the quantum mechanical magnetic properties of an atom's nucleus' (Nuclear Magnetic Resonance in Imaging Techniques - Wikipedia, the free encyclopedia, 2009).

All nuclei that contain odd numbers of protons or neutrons have both an intrinsic magnetic momentum (Magnetic momentum in Physics - Wikipedia, the free encyclopedia, 2009) and an angular momentum (Angular momentum in Physics - Wikipedia, the free encyclopedia, 2009).

The most commonly used nuclei are *hydrogen-1* (the most receptive isotope at natural abundance; see also par. 3.2 below) and *carbon-13*.

2.2. Definition

In the 1940-1950 years, Bloch and Purcell noticed that magnetic nuclei, like hydrogen (H) and potassium (P), could absorb RF (Radiofrequencies Fields) energy when placed in a magnetic field of a strength specific to the identity of the nuclei. When this absorption occurs, the nucleus is described as being *in resonance*: different atoms within a molecule *resonate* at different frequencies at a given field strength. The observation of the resonance frequencies of a molecule allows a scientist to discover structural information about the molecule. NMR resonant frequencies for a particular substance are directly proportional to the strength of the applied magnetic field, in accordance with the equation for the Larmor precession frequency (Larmor frequency in Physics - Wikipedia, the free encyclopedia, 2009).

Nowadays, Nuclear Magnetic Resonance has been defined as a phenomenon which occurs when the nuclei of certain atoms are immersed in a static magnetic field and exposed to a second oscillating magnetic field. Some nuclei experience this phenomenon, and others do not, dependent upon whether they possess a property called ‘spin’ (Hornak, 1997-1999).

In other words, when the nuclear magnetic momentum associated with a nuclear spin is placed in an external magnetic field, the different spin states are given different magnetic potential energies. In the presence of the static magnetic field which produces a small amount of spin polarization, a radiofrequency signal of the proper frequency can induce a transition between spin states. This ‘spin flip’ places some of the spins in their higher energy state. If the radiofrequency signal is then switched off, the relaxation of the spins back to the lower state produces a measurable amount of RF signal at the resonant frequency associated with the ‘spin flip’ (Nuclear Magnetic Resonance in Imaging Techniques - Wikipedia, the free encyclopedia, 2009) (Figure 15.1).

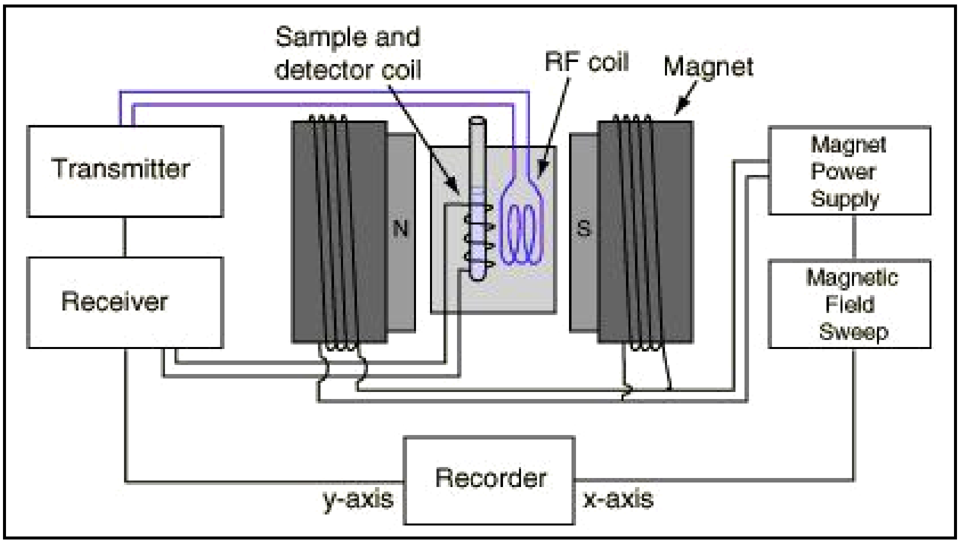


Figure 15.1. Functional principles of the Nuclear Magnetic Resonance.

NMR studies magnetic nuclei by aligning them with an applied constant magnetic field and perturbing this alignment using an alternating magnetic field. Particularly, those fields have to be orthogonal. The resulting response to the perturbing magnetic field is the phenomenon that is exploited in a) NMR spectroscopy; and b) Magnetic Resonance Imaging (see par. 3 below), which use very powerful applied magnetic fields in order to achieve high spectral resolution.

2.3. Sensitivity

Because the intensity of NMR signals and, hence, the sensitivity of the technique depends on the strength of the magnetic field, the technique has also advanced over the decades with the development of more powerful magnets (Nuclear Magnetic Resonance in Imaging Techniques - Wikipedia, the free encyclopedia, 2009). Focusing on the sensitivity of NMR signals, makes it clear that this sensitivity is dependent on: a) the presence of a magnetically-susceptible nuclei; b) the natural abundance of such nuclei; c) the ability of the scientist to artificially enrich the molecules with such nuclei; d) the quantum-mechanical nature of the phenomenon; e) the temperature; and f) the saturation of the sample with energy applied at the resonant radiofrequency.

2.4. Theory of Nuclear Magnetic Resonance

2.4.1. Introduction: nuclear spin and spin angular momentum

The elementary particles (neutrons and protons), composing an atomic nucleus, have the intrinsic quantum mechanical property of spin (Spin in Physics - Wikipedia, the free encyclopedia, 2009). Particularly, the overall spin of the nucleus is determined by the spin quantum number I (Bloch, 1946).

a) If the number of the protons and neutrons in a given isotope is even then I = 0, just as electrons pair up in atomic orbits, so do even numbers of protons and neutrons pair up giving zero overall spin.

b) In other cases, however, the overall spin is non-zero.

Consequently, a non-zero spin I is associated with a non-zero magnetic momentum μ:

$$\mu = \gamma \bullet I \quad \text{[Eq. 15.1]}$$

where the proportionality constant γ is the gyro-magnetic ratio. It is this magnetic momentum that is exploited in NMR (Figure 15.2).

Note that the electron spin magnetic momentum is opposite to the electron spin while the proton spin magnetic momentum is in the direction of the proton spin. The basic principles are otherwise similar.

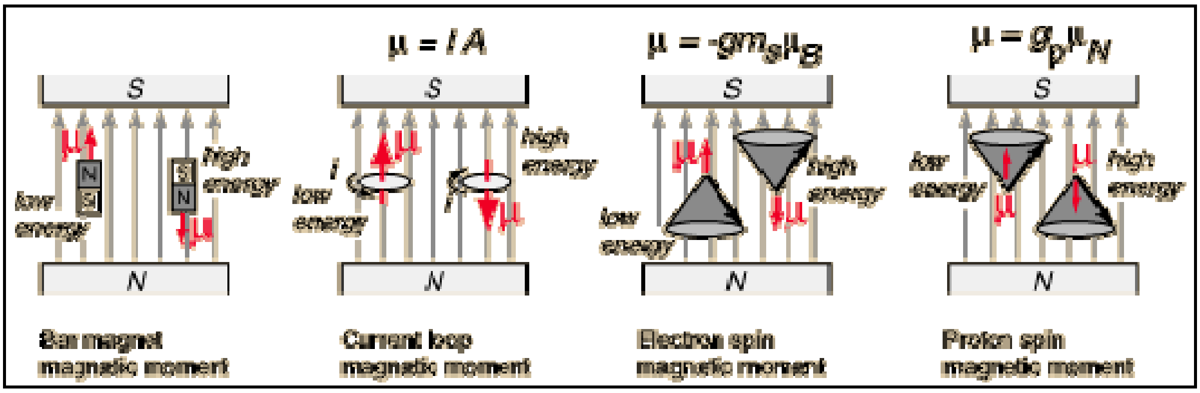
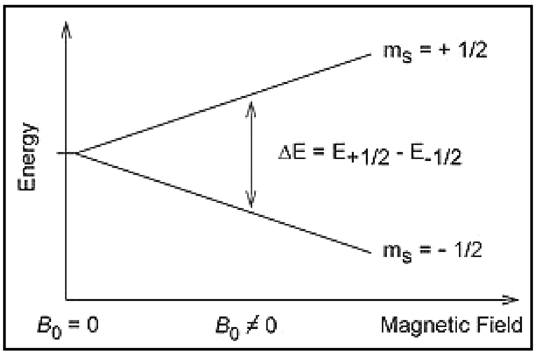


Figure 15.2. The pattern of a magnetic momentum in a magnetic field.

Resonant absorption will occur when electromagnetic radiation of the correct frequency to match the energy difference between the nuclear spin levels in a constant magnetic field of the appropriate strength is applied. These frequencies typically correspond to the radiofrequency range of the electromagnetic spectrum (Larmor frequency in Physics - Wikipedia, the free encyclopedia, 2009). It is this resonant absorption that is detected in NMR. Finally, the angular momentum μ associated with nuclear spin is quantized, meaning both that the magnitude of angular momentum is quantized and also that the orientation of the associated angular momentum is quantized (Angular momentum in Physics - Wikipedia, the free encyclopedia, 2009).

2.4.2. The behavior of spin in a magnetic field

In general, a nucleus has two possible spin states: +½ or -½ (also referred to as up and down or α and β, respectively; Figure 15.3. Spin states in a magnetic field).



The energies of these states are degenerate. In other words, they are the same (Figure 15.4, left graph). Hence the populations of the two states (i.e. number of atoms in the two states) will be approximately equal at thermal equilibrium.

Furthermore, if a nucleus is placed in a magnetic field (Figure 15.4, right graph), the interaction between the nuclear magnetic momentum and the external magnetic field means the two states no longer have the same energy. In fact, the energy of the magnetic momentum μ when in a magnetic field B_0 is given by the negative scalar product of the vectors. As a result, the different nuclear spin states have different energies in a non-zero magnetic field. In hand-waving terms, we can talk about the two spin states of a spin ½ as being aligned either with or against the magnetic field. If γ is positive (true for most isotopes) then m = ½ is the lower energy state.

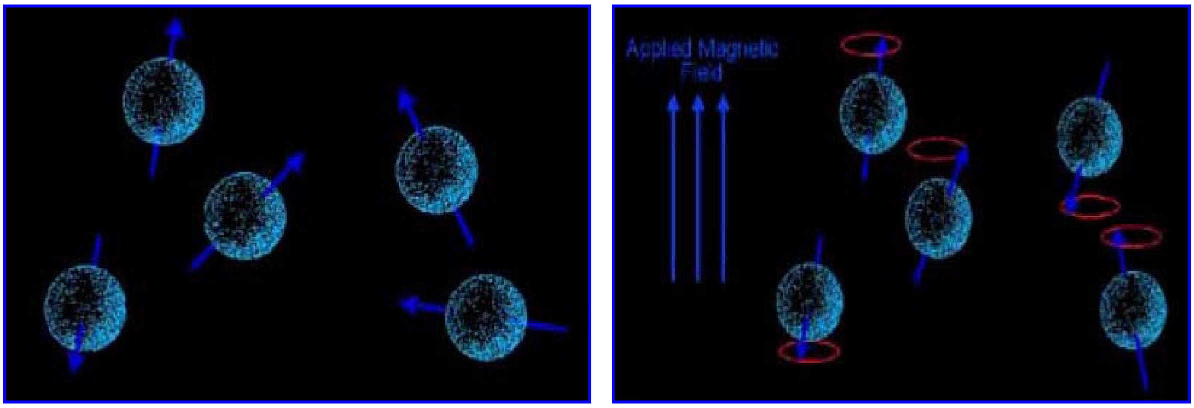


Figure 15.4. Nucleus non-placed (on the left) and placed (on the right) in a magnetic field.

2.4.3. Nuclear shielding

It might appear that all nuclei of the same nuclide (and hence the same γ) would resonate at the same frequency. However, this is not the case. Particularly, the most important perturbation of the NMR frequency is the *shielding effect* of the surrounding electrons (Raynes, 1972). In general, this electronic shielding reduces the magnetic field at the nucleus (which is what determines the NMR frequency). As a result, the energy gap is reduced and the frequency required to achieve resonance is also reduced. Unless the local symmetry is particularly high, the shielding effect depends on the orientation of the molecule with respect to the external field.

2.4.4. Relaxation

The process called *relaxation* refers to nuclei that return to the thermodynamic state in the magnet (Bloemberger et al., 1948). In particular:

- this process is also called ***T1 relaxation***, where T1 refers to the mean time for an individual nucleus to return to its equilibrium state. Moreover, once the population is relaxed, it can be probed again, since it is in the initial state;
- the precessing nuclei (see par. 2.2 above) can also fall out of alignment with each other and stop producing a signal. This is called ***T2 relaxation***. It is possible to be in this state and not have the population difference required to give a net magnetization vector at its thermodynamic state.

T1 is always larger (and slower) than T2: this happens because some of the spins were flipped by the pulse. Indeed, they will remain so until they have undergone population relaxation.

In practice, the value of T2* which is the actually observed decay time in the NMR signal, or free induction decay, also depends on the static magnetic field in-homogeneity, which is quite significant.

2.5. Application fields

The versatility of NMR makes it pervasive in the sciences. Scientists and students are discovering that knowledge of the science and technology of NMR is essential for applying, as well as developing, new applications for it (Hornak, 1997-1999; Tyszka et al., 2005).

Specifically, NMR has been developed in many fields over the last years:

a. Chemistry. By studying the peaks of NMR spectra, chemists could determine the structure of many compounds (in terms both of dynamics and molecular motion; Wuthrich, 1986), also comparing the observed nuclear precession frequencies to known frequencies;

b. Non-destructive testing. NMR is extremely useful to analyze samples non-destructively because radio waves and static magnetic fields easily penetrate many types of matter and anything that is not inherently ferromagnetic;

c. Data acquisition in the petroleum industry. NMR analysis of boreholes is used to measure rock porosity, estimate permeability from pore size distribution and identify pore fluids;

d. Process control. NMR has now entered the arena of real-time process control and process optimization in oil refineries and petrochimical plants;

e. Quantum computing, using the spin states of molecules. Specifically, NMR differs from other implementations of quantum computers in that it uses an ensemble of systems, in this case molecules. The ensemble is initialized to be the thermal equilibrium state;

f. Magnetic Resonance Imaging. NMR is widely used both in chemical studies (MRI spectroscopy) and in medical diagnosis (Magnetic Resonance Imaging; see par. 3 below). These studies are possible because nuclei are surrounded by orbiting electrons, which are also spinning charged particles such as magnets and, so, will partially shield the nuclei with respect to the exact local environment.

In the next sections, we will briefly focus on the medical technique of Magnetic Resonance Imaging (MRI), because it was used to record static symmetries.

3. MAGNETIC RESONANCE IMAGING

3.1. Definition

With recent technologic advances, including faster acquisition times and better anatomic and pathological depiction, Magnetic Resonance Imaging (MRI) is used with increasing frequency (Alterson et al., 2003; Bitar et al., 2006).

In detail, MRI is primarily a medical imaging technique most commonly used in radiology to visualize the structure and function of the body. In fact, it provides detailed images of the body in any plane (Mattson et al., 1996; Brown et al., 1999; Mündermann et al., 2006; Neu et al., 2009; Racic et al., 2009; Riad et al., 2009).

It was developed from knowledge gained in the study of NMR (see par. 2.1 onwards). In its early years, the technique was referred to as Nuclear Magnetic Resonance Imaging (NMRI). However, as the word *nuclear* was often associated with ionizing radiation exposure, it is generally now referred to simply as MRI (Brown et al., 1999). The typical MRI examination consists of 5-20 sequences, each of which are chosen to provide a particular type of information about the subject tissues. This information is then synthesized by the interpreting physician.

Unlike Computed Tomography (CT), MRI uses no ionizing radiation but a powerful magnetic field to align the nuclear magnetization of (usually) hydrogen atoms in water in the body (Deck et al., 1989; Brown et al., 1999; Bitar et al., 2006; Warwick et al., 2009). Radiofrequency fields are used to systematically alter the alignment of this magnetization, causing the hydrogen nuclei to produce a rotating magnetic field detectable by the scanner. This signal can be manipulated by additional magnetic fields to build up enough information to construct the last image of the body.

MRI is used to image every part of the body (Grossman et al., 2008), but is particularly useful in neurological conditions, disorders of the muscles and joints (Fleckenstein et al., 1991), in investigating muscle activity (Roberts et al., 2005), for evaluating tumors and showing abnormalities in the heart and blood vessels and in assessing tissue mechanical function (Colosimo et al., 2005; Golder, 2007; Neu et al., 2009; Riad et al., 2009).

3.2. Main physical principles

MRI is based on 'the electromagnetic activity of atomic nuclei' (Bloch, 1946). Nuclei are made up of protons and neutrons, both of which have spins. MR-active nuclei are those that have a net spin because they are odd-numbered and the spins of their protons and neutrons do not cancel each other out (Brown et al., 1999; Bitar et al., 2006). ^1H is the most commonly nucleus used because it seems very useful and highly abundant.

Particularly, the human body is mainly composed of water molecules (≈ 90%; Jackson et al., 1985; Cunnigham et al., 1991; Heymsfield et al., 1997; McArdle et al., 2001) which each contain two hydrogen nuclei (or protons). Each nucleus rotates around its own axis. As the nucleus spins, its motion induces a magnetic field. When the nuclei are exposed to an external magnetic field (B_0), the interaction of the magnetic fields causes the nuclei to wobble or precess (Bloch, 1946).

The frequency at which precession occurs is defined by the Larmor equation (see par. 2.2 and 2.4.1 above):

$$\omega_0 = B_0 \cdot \gamma \text{ [Eq. 15.2]}$$

where ω_0 is the precessional frequency; B_0 the external magnetic field strength (in Tesla; Figure 15.6 below. External magnetic field); and γ the gyro-magnetic ratio (in megahertz *per* Tesla), which is a constant for every atom at a particular magnetic field strength.

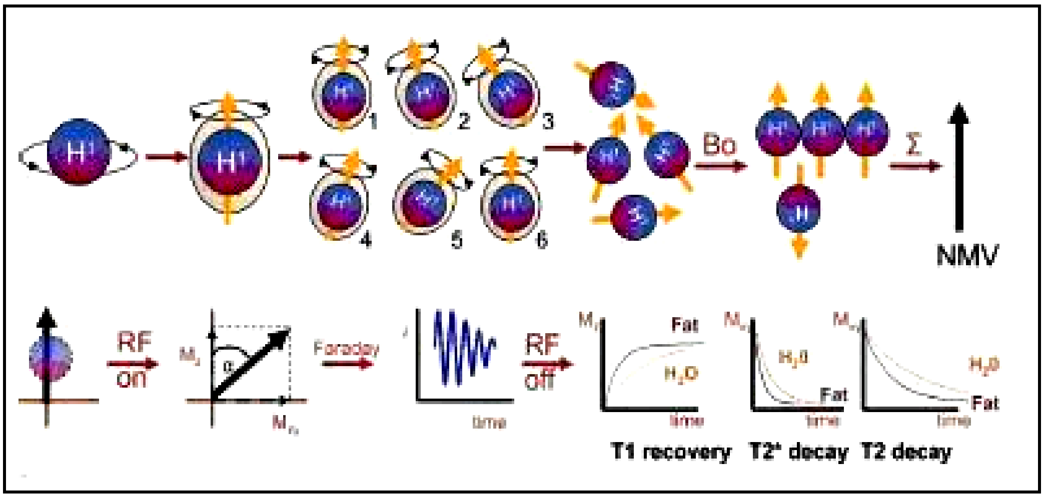
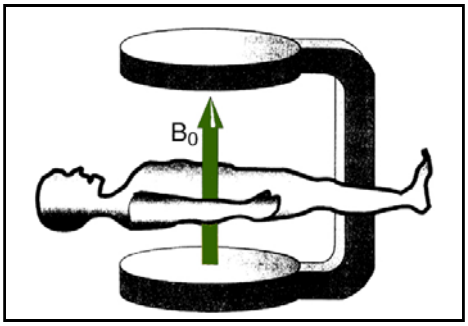


Figure 15.5. Basic physics of the MR signal, in Bitar et al. (2006).



Until the hydrogen nuclei are exposed to B_0 magnetization, their axes are randomly aligned. However, when B_0 magnetization is applied, the magnetic axis of the nuclei align with the magnetic axis of B_0, some in parallel and other in opposition to it (Figure 15.5 above). The cumulative effect of all the magnetic moments of the nuclei is 'the net magnetization vector'.

When a radiofrequency (RF) pulse is applied, the RF excitation causes the net magnetization vector to flip a certain angle, and this produces two magnetization vector components: 1) longitudinal magnetization and 2) transverse magnetization. As the transverse magnetization precesses around a receiver coil, it induces a current in that coil. This current becomes the MR signal (Brown et al., 1999). Furthermore, when the RF energy source is turned off, the net magnetization vector realigns with the axis of B_0 through the precess of (Bitar et al., 2006):

a) the recovery of longitudinal magnetization (called ***longitudinal*** or ***T1 relaxation***; Figure 15.7, left graph). It occurs exponentially with the time constant T1 and it is due to a very little excess of protons in the lower energy state;

b) the loss of phase coherence in the transverse plane (called ***transverse*** or ***T2 relaxation*** or ***T2* decay***; Figure 15.7, right graph). It is due to coherences forming between the two proton energy states.

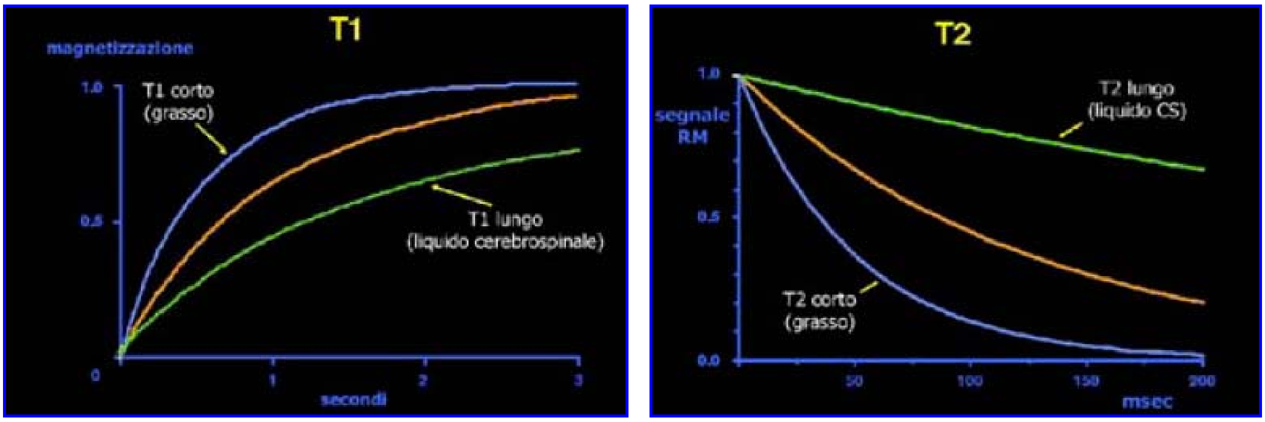


Figure 15.7. Longitudinal/T1 relaxation (on the left) and transverse/T2 relaxation (on the right).

Different tissues have different T1, T2 and T2* values. Furthermore, T2* is dependent on the magnetic environment (Bitar et al., 2006). Particularly:

a) T1 is associated with the *enthalpy* (Enthalpy in Biochemistry - Wikipedia, the free encyclopedia, 2009) of the spin system. Fat has a shorter T1 (i.e. recovers faster); indeed, water has a relatively long T1 (Figure 15.5 above). During T1 relaxation, the longitudinal magnetization recovers as the spinning nuclei release energy into the environment (Figure 15.8 below);

b) T2 is associated with the *entropy* system (Pincus, 1991; Entropy in Biochemistry - Wikipedia, the free encyclopedia, 2009) of the spin. Fat has a shorter T2 (i.e. recovers faster); indeed, water has a relatively long T2 (Figure 15.5 above). During T2 relaxation, the transverse magnetization is de-phased because of interaction between the spinning nuclei and their magnetic fields (Figure 15.8 below);

c) T2* decay occurs very quickly both in fat and water (Figure 15.5 above). In T2* decay, the transverse magnetization is de-phased because of magnetic field in-homogeneities (Figure 15.8 below).

As shown in Figure 15.5 above, typically in soft tissues, T1 is around one second, while T2 is a few tens of milliseconds; however, these values vary widely between different tissues (and different external magnetic fields), giving MRI its tremendous soft tissue contrast. Contrast agents work by altering (shortening) the relaxation parameters, especially T1, too (Bloemberger et al., 1948).

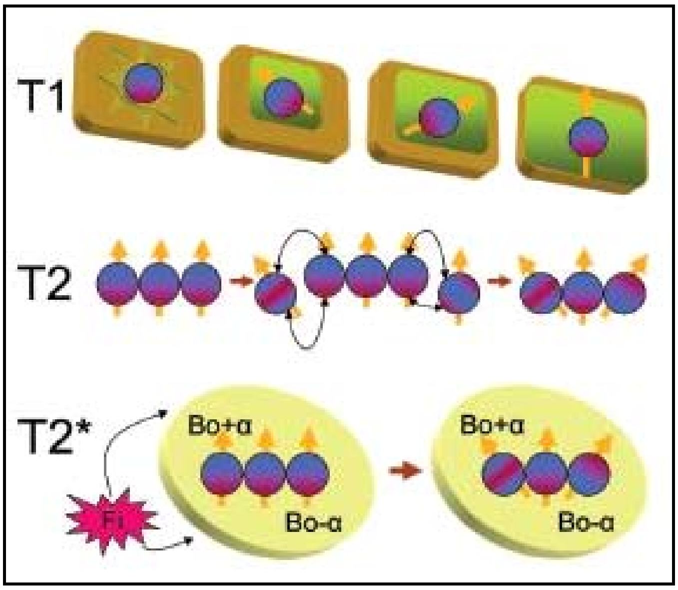


Figure 15.8. Magnetization relaxation and decay, in Bitar et al. (2006).

The process, in which the transverse component of the net magnetization vector induces a current in the receiver coil (Brown et al., 1999; Bitar et al., 2006), could be called *free* (e.g. refers to the fact that the system in no longer being forced out of equilibrium by the RF excitation) *induction* (e.g. describes the mechanism through which the signal is detected) *decay* (e.g. refers to the decrease in signal amplitude over time).

To sum up, if a person goes inside the powerful magnetic field of a scanner (Figure 15.9, left graph), these protons align with the direction of the field (Figure 15.9, middle graph). A second radiofrequency electromagnetic field is then briefly turned on causing the protons to absorb some of its energy. If this field is turned off, the protons release this energy at a radiofrequency which can be detected by the scanner (Figure 15.9, right graph).

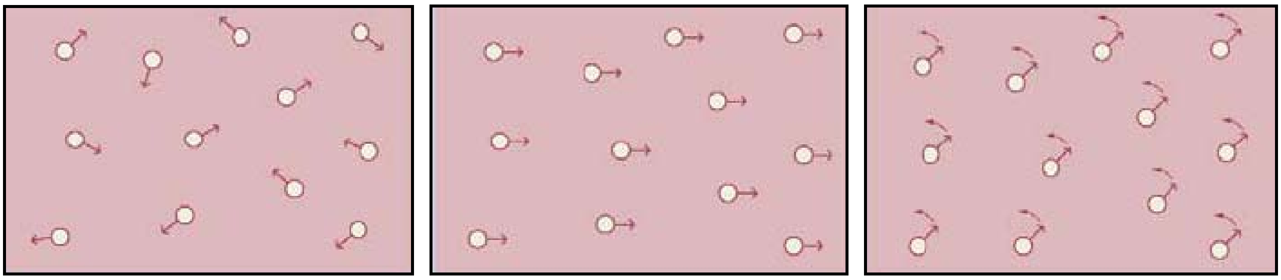


Figure 15.9. Protons pattern in a normal state (on the left), in a magnetic field (in the middle) and after the application of a radiofrequency (on the right).

Consequently, the position of protons in the body can be determined by applying additional magnetic fields (and by turning gradients coils on and off) during the scan which allows an image of the body to be built up. In other words, when certain nuclei such as H (Hydrogen), Na (Sodium) or P (Potassium) with their spins are placed in a strong external magnetic field, they precess around an axis along the direction of the field. Particularly, in the static magnetic fields commonly used in MRI, the energy difference between the nuclear spin states corresponds to a photon at radiofrequency wavelengths. Moreover, resonant absorption of energy by the protons due to an external oscillating magnetic field will occur at the Larmor frequency for each particular nucleus.

3.3. Imaging technique: scheme and contrast

To combine field gradients and radiofrequency excitation in order to create an image, some different schemes have been devised for: a) some of them involve 2D or 3D reconstruction from projections; b) some involve building the image point-by-point or line-by-line; and c) others even use gradients in the RF field rather than the static field.

Although each of these schemes has been occasionally used in specialist applications, the majority of MR Images today are created either by the two-dimensional Fourier Transform (2DFT) technique with slice selection or by the three-dimensional Fourier Transform (3DFT) technique. For more details about this argument, see also Lauterbur (1973), Kumar et al. (1975), Ridway et al. (1986), Hawkes et al. (1987) and Brown et al. (1999).

Image contrast is then created by differences in the strength of the NMR signal recovered from different locations within the sample. This depends on (Bitar et al., 2006): a) the relative density of excited nuclei (usually water protons); and b) differences in relaxation times (T1 and T2) of those nuclei after the pulse sequence.

Two parameters are key to the creation of image contrast:

- Time Repetition (TR). It is the time between the application of an RF excitation pulse and the start of the next RF pulse. TR relates to T1 and affects contrast on T1-weighted images (Figure 15.5 above and Figure 15.10 below);
- Time Echo (TE). It is the time between the application of the RF pulse and the peak of the echo detected. TE relates to T2 and affects contrast on T1-weighted images (Figure 15.5 above and Figure 15.10 below).

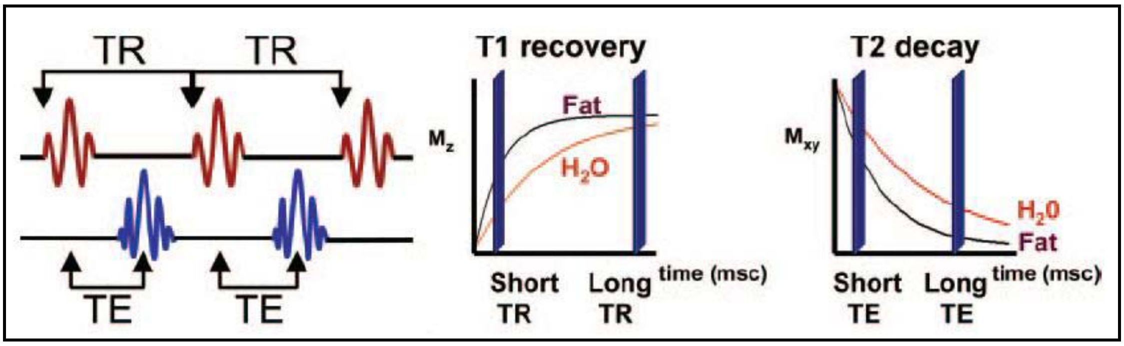


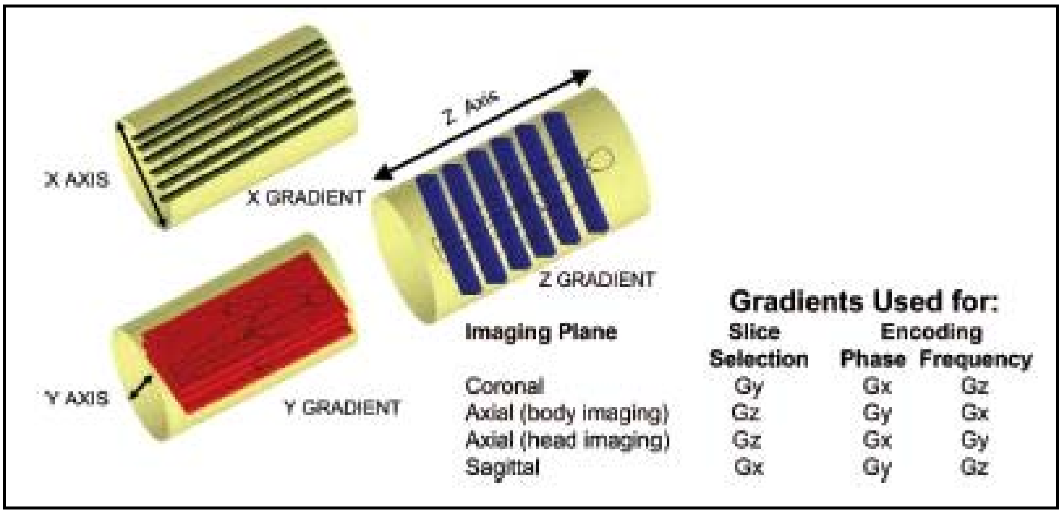
Figure 15.10. Schematic representation of RT and ET, in Bitar et al. (2006).

In most MR images, contrast is actually a mixture of all these effects (Bitar et al., 2006).

Therefore, the ability to choose different contrast mechanisms gives MRI tremendous flexibility (e.g. in the brain, T1-weighting causes the nerve connections of white matter to appear white, whereas the congregations of neurons of grey matter to appear grey and cerebrospinal fluid (CSF) appears dark). Finally, in some situations it is not possible to generate enough image contrast to adequately show the anatomy or pathology of interest by adjusting the imaging parameters alone, in which case a contrast agent may be administered (i.e. paramagnetic contrast agent, super-paramagnetic contrast agent and diamagnetic agent).

How does the MR imaging system detect which tissue the signal is coming from?

In order to know this relevant information, gradients have to be employed (Bitar et al., 2006). Usually, three type of gradients are applied, respecting the axis of imaging (Figure 15.11).



Imaging Plane	Gradients Used for: Slice Selection	Encoding Phase	Encoding Frequency
Coronal	Gy	Gx	Gz
Axial (body imaging)	Gz	Gy	Gx
Axial (head imaging)	Gz	Gx	Gy
Sagittal	Gx	Gy	Gz

Figure 15.11. Schematic x, y and z axis gradients, in Bitar et al. (2006).

The section-selective gradient selects the section to be imaged.

Of utmost importance, pulse sequences are wave forms of the gradients and RF pulses (Bitar et al., 2006), applied in MR image acquisition. Consequently, the pulse sequence diagram could be considered as a schema of the timing of instructions sent to the RF generator and gradient amplifiers.

There are two fundamental types of MR pulse sequences (two-dimensional or three-dimensional): SE and GRE. All other MR sequences are variation of these, with different parameters added on. For other information about these two types, we could refer to Bitar et al. (2006).

Finally, it is important that technicians and radiologists who work in MR imaging be acquainted with the concept of the specific absorption rate (SAR). The SAR is ‘a measure of the rate at which that RF energy (in watts) is dissipated in tissues *per* unit of tissue mass’ (in kilograms; Bitar et al., 2006).

3.4. Main advantages

The main advantages of using MRI are that, in clinical practice (Mirarchi, 2007):

- it is not so expensive;
- it is harmless to the patient;
- it uses strong magnetic fields and no-ionizing radiation in the radiofrequency range;
- no effects of MRI on the fetus have been demonstrated so that it is rapidly growing in importance as a way of diagnosing and monitoring congenital defects of the fetus;
- it is able to distinguish pathologic tissue from normal one;
- it provides comparable resolution with far better contrast resolution;
- modern scanners may have larger bores and scan times are shorter in order to avoid handicap of claustrophobia or other discomforts.

3.5. Instrumental description

The main components (Seminati et al., 2001) necessary to make a MRI could be summarized in:

a) a magnet. It constitutes the main element of the MRI system, generating a static magnetic field. Its properties depend on the intensity and the homogeneity of this field. In theory, three different groups of magnets could be used: permanent magnets (e.g. they are able to keep a low intensity magnetization over a long time without requiring an external device), resistive magnets (e.g. they work on the flow of a continuous current producing a moderate intensity magnetization), and super-conductive magnets (e.g. they work on a super-conductive current generating a high intensity magnetization);

b) coils and gradient generators (see par. 3.3 above);

c) a transmitter that generates the radiofrequency pulses to excite nuclei and a frequency receiver that receives and converts the pulses into an electric signal;

d) a calculus system that controls, acquires and reconstructs the MR images;

e) a station operator which permits the technician to check and monitor all exam phases operating to the calculus system by suitable TV monitor;

f) a patient bed without any ferromagnetic objects.

The instrumentation we used to acquire a MRI has been represented in Figure 15.12.

Figure 15.12. The instrumentation we used to acquire a Magnetic Resonance Imaging.

3.6. Application fields

Because of its peculiar efficiency, MRI has been widely used in many different fields:

a. Diffusion MRI. It measures both the diffusion of water molecules in biological tissues and the diffusion-weighted imaging making brain maps of fiber directions to examine the connectivity of different regions in the brain or to examine areas of neural degeneration;

b. Magnetization Transfer. It refers to the transfer of longitudinal magnetization from free water protons to hydration water protons in NMR and MRI;

c. Fluid Attenuated Inversion Recovery. It is a pulse sequence used in MRI to null fluids;

d. Magnetic Resonance Angiography. It is used to generate pictures of the arteries in order to evaluate them both for stenosis or aneurysms;

e. Magnetic Resonance Spectroscopy. It is used to measure the levels of different metabolites in body tissues and it combines both spectroscopic and imaging methods to produce spatially localized spectra from within the sample or patient;

f. Functional MRI. It measures signal changes in the brain that are due to changing neural activity;

g. Interventional MRI. It is often used to guide minimally-invasive procedures;

h. Radiation Therapy Simulation. It is now being utilized to specifically locate tumors within the body in preparation for radiation therapy treatments;

i. Current Density Imaging. It endeavors to use the phase information from images to reconstruct current densities within a subject;

j. Multinuclear Imaging Hydrogen. It is the most frequently imaged nucleus in MRI because it is present in biological tissues in great abundance. However, any nucleus which has a net nuclear spin could potentially be imaged with MRI (Table 15.1);

NUCLEI	IMPAIRED	SPIN
^{1}H	1	1/2
^{2}H	1	1
^{31}P	0	1/2
^{23}Na	0	3/2
^{14}N	1	1
^{13}C	0	1/2
^{19}F	0	1/2

Table 15.1. The most frequently imaged nucleus in MRI.

k. Susceptibility Weighted Imaging. It is a new type of contrast in MRI different from spin density, T1 or T2 imaging;

l. Measurements of the segment inertial properties (Martin et al., 1989; Mungiole et al., 1990).

4. PORTABLE METABOGRAPH

4.1. Introduction

With advances in computer and microprocessor technology, the exercise scientist can measure metabolic and physiologic responses to exercise accurately and rapidly (McArdle et al., 2001).

In fact, devices to measure energy expended both at rest and whilst performing particular activities have been devised (Littlewood et al., 2002). In general, a computer interfaces with:

a) a system to continuously sample the subject's expired air;

b) a flow-measuring device to record breathed air volume;

c) oxygen and carbon dioxide analyzers to measure the expired gas mixture's composition.

One of these more-advanced devices/systems is the $K4b^2$. It is the first Cosmed portable system for intrapulmonary gas exchange analysis on real breath by breath basis (*indirect calorimetry*; Bar-Haim et al., 2008; $K4b^2$ in Equipment for exercise physiology - Cosmed, 2008; Brouwer et al., 2009; McGregor et al., 2009). Particularly, it is a portable, lightweight telemetric system that measures the volume of expired air, volume of oxygen consumed in litres *per* minute, volume of carbon dioxide expired in litres *per* minute, heart rate and respiratory frequency (Littlewood et al., 2002). Its technology and dimensions allow scientists to measure physiological responses to exercise (also in the field) without any limits.

K4b^2 was designed to be accurate, reliable and valid (Fleisch, 1953; Kawakami et al., 1992; Lothian et al., 1993; Lucia et al., 1993; Peel et al., 1993; Corry et al., 1996; Pinnington et al., 2001; Duffield et al., 2004; K4b^2 Manual, 2004) in any conditions (Bosco et al., 1997; Hausswirth et al., 1997; Jensen et al., 1999; Saunders et al., 2004; McNaughton et al., 2005; Minetti et al., 2006; Zamparo et al., 2008; Brouwer et al., 2009; Lepretre et al., 2009).

The fast CO_2 and O_2 analyzer is maintained at a constant temperature. Sampling flow and pressure are continuously monitored. A barometer along with a temperature and pressure sensor allow instantaneous correction in any change in the environmental conditions. As a result, K4b^2 is a versatile system. Whether in the laboratory or in the field, tests can be carried out in three different configurations: 1) holter data recorder; 2) telemetry transmission; and 3) laboratory station (K4b^2 Manual, 2004).

4.2. The main components

K4b^2 main components are:

- face mask (Figure 15.13: right graph), allowing the sampling of the expired air and the photoelectric turbine which measures minute ventilation (Kawakami et al., 1992; Littlewood et al., 2002; Mian et al., 2006);
- control unit (portable), representing the main unit of the system (Figure 15.13: left graph). Particularly, it contains both the oxygen and carbon dioxide sensors, sampling pump, barometric sensors and electronics (Kawakami et al., 1992; Littlewood et al., 2002). Moreover, it permits scientists to recorder anthropometric measurements of the subject, to start and stop (and save) a test and to calibrate the system (see par. 4.3 below);

Figure 15.13. Portable metabograph K4b^2 (Cosmed, Italy).

- bidirectional turbine, regulating the gas exchange (Figure 15.14);

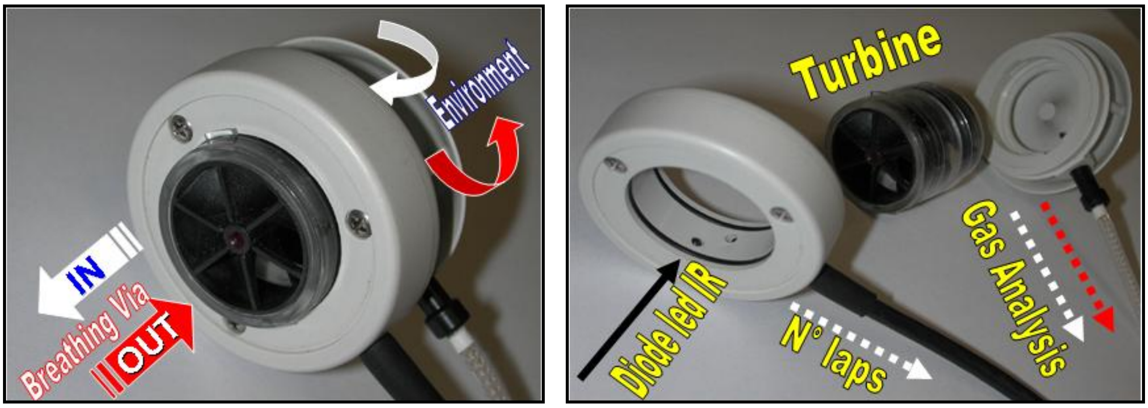


Figure 15.14. Bidirectional turbine of the portable metabograph K4b^2.

- receiving station (or electrode; Kawakami et al., 1992; Littlewood et al., 2002) for heart rate;
- eventual receiving station GPS;
- telemetry antenna.

4.3. Calibration procedure

A subject places the heart rate monitor around his/her chest (Littlewood et al., 2002).

The system has to be switched on at least 45 minutes before starting the calibration procedure.

Calibration of the Cosmed K4b^2 is carried out prior to the measurement of a subject (Corry et al., 1996). The calibration of the metabograph constitutes the most important phase: the good acquisition of data (and relative values) depends on this phase.

It is possible to calibrate the system directly on the portable unit or via software. In our experiments, the system was calibrated directly on the portable unit, in the Biomechanics Laboratory of the Faculty of Exercise and Motor Science where tests have been carried out.

Specifically, we will briefly focus on this important procedure. It is possible to perform the following steps in any order (Littlewood et al., 2002): a) room air calibration. This updates the baseline of the main values of gas (O_2 analyser and CO_2 analyser) in room air. In this way, the gases would be in line with predicted values within the atmosphere (20.93% for O_2; 0.03% for CO_2; McGregor et al., 2009); b) turbine calibration, by the proper syringe that permits to introduce a known volume (3 l) inside the sampling unit (Bar-Haim et al., 2008); c) reference gas calibration. It permits to record the main values of gas (O_2, CO_2) in a known composition of a gas cylinder (16% for O_2; 5% for CO_2; McGregor et al., 2009); and d) delay calibration, connecting the mask to the sampling unit. It permits to synchronize the human breath with the parameters of the metabograph (Littlewood et al., 2002).

The subject needs to wear the heart rate belt, the mask and the 'jacket' in which the portable unit (front) and the battery (back) are fixed. To avoid interferences, the heart rate antenna has to be fixed on the belt just below the subject's left shoulder; all the other wires need to be fixed on the belt just below the subject's right shoulder, i.e. using velcro.

In our experiments, data sent via telemetry to a computer. Once the air calibration has been carried out, the system begins to record and the clock/timer stops blinking. At the end of the recording session, it is important to disinfect and dry all the instruments the subject put on.

In pictures at the end of the chapter, some snapshots (Figure 15.15: V'E, HR and $V'CO_2/V'O_2$ *versus* time) from $K4b^2$ software we used (version 7.8) are proposed.

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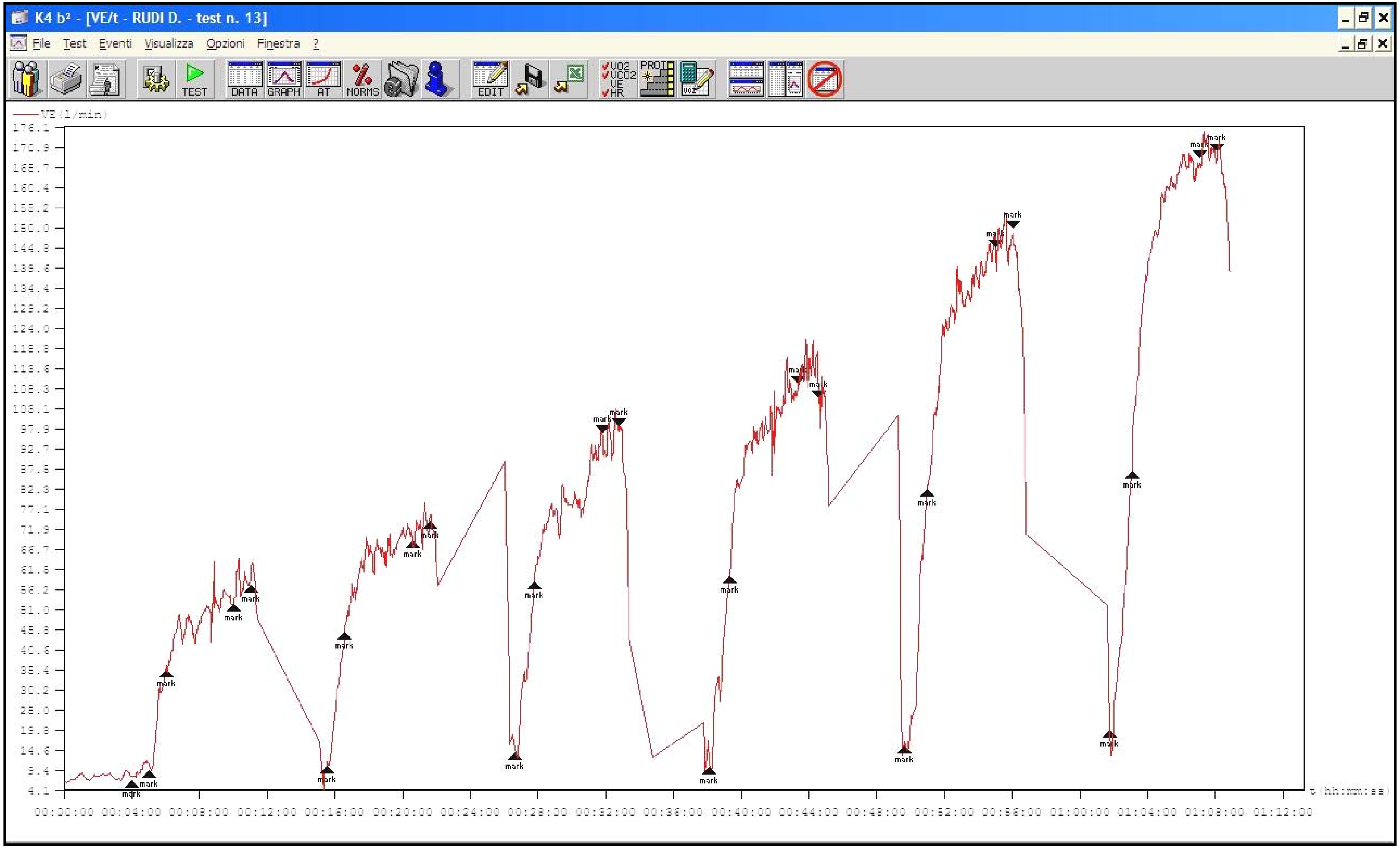
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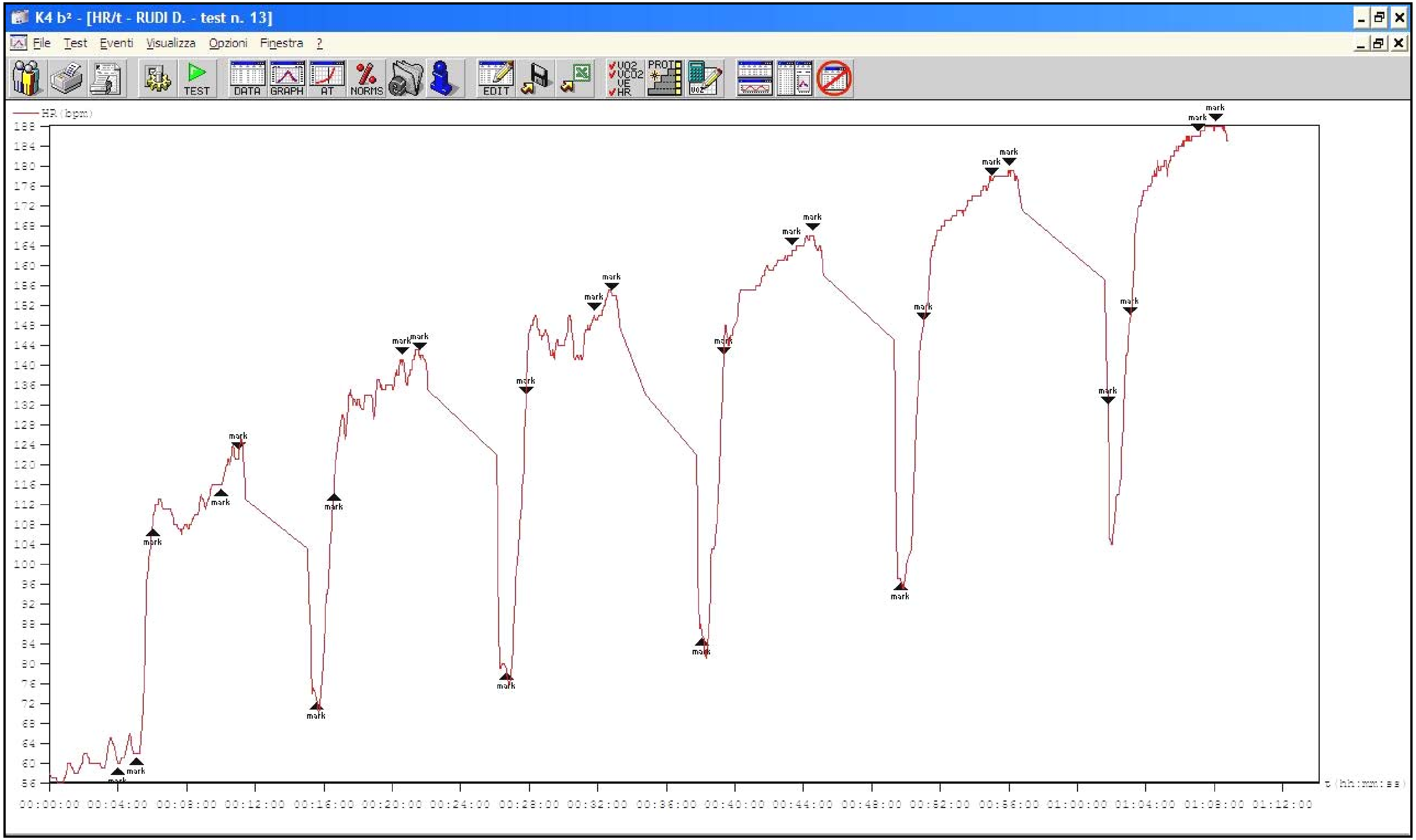
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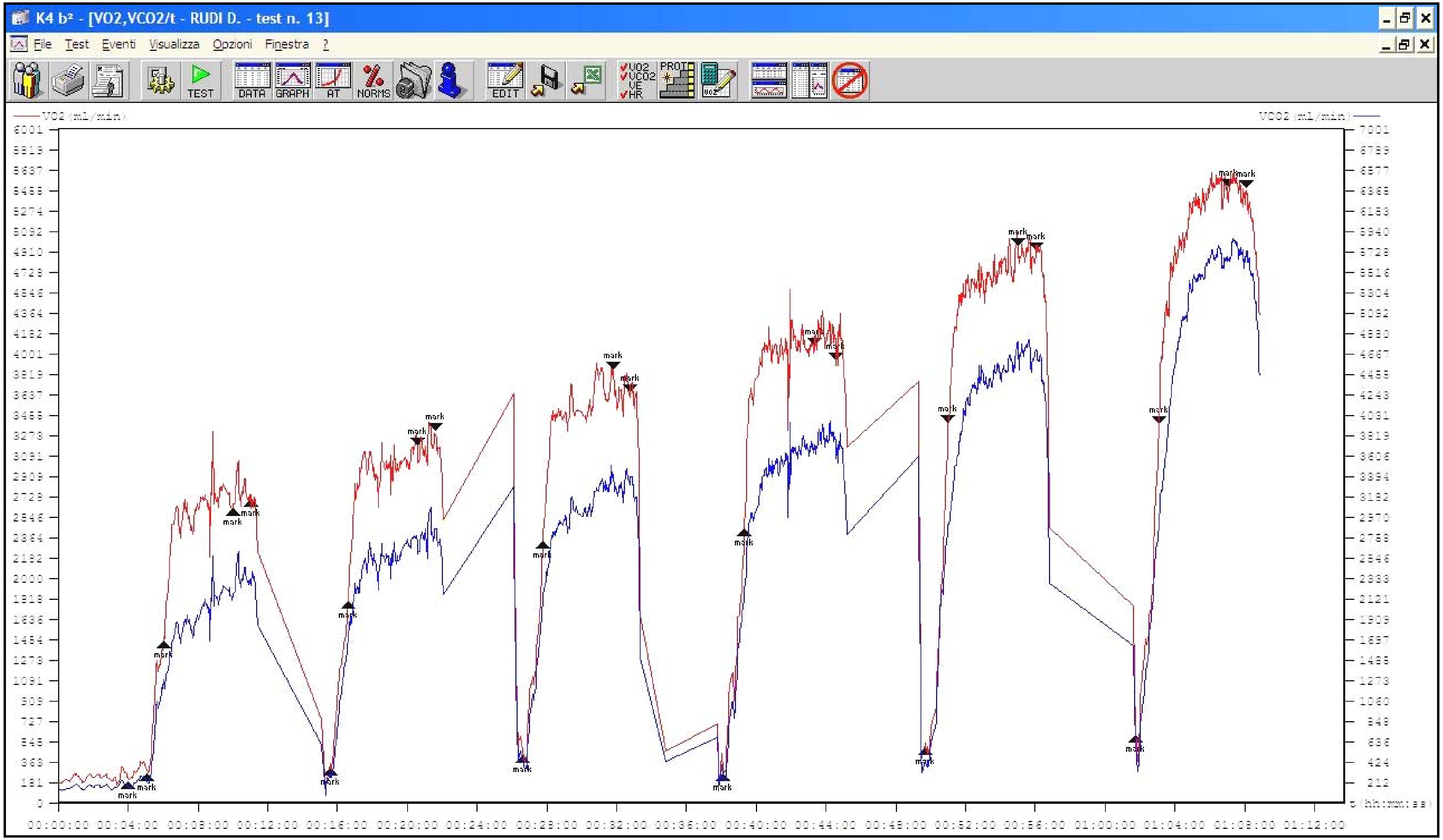

Figure 15.15. Graphical snapshots of a trial running in our experiments.

Chapter 16

SUBJECTS

1. RUNNERS

We were interested in healthy male subjects of different ages (from 20 to 55 years), featuring different running abilities (Brisswalter et al., 1994; Lees et al., 1994; Morgan et al., 1995; Pereira et al., 1997; Kyrolainen et al., 2000; Kilding et al., 2005; Marcovic et al., 2007; Manzi et al., 2009; McGregor et al., 2009). Consequently, we created three test groups (7 people *per* group in OR and SR; 5 people in TR), based on this specific running ability:

- **occasional runners** (OR). They normally do sport (e.g. football, athletics) but not specifically running (training less than 2 hours *per week*). As a result, they are moderately active people: they practice sport 3 times *per week*. Their best marathon time (BMT) is above 5 hours;
- **skilled runners** (SR). They are in prevalence triathletes (Millet et al., 2003) who have taken part in at least two marathons and they are highly fit athletes who train more than 3 times *per week* (training from 2 to 6 hours *per week*). Their BMT ranges from 3 to 5 hours;
- **top runners** (TR). They have taken part in at least two marathons and they are highly fit athletes who train more than 3 times *per week* (training 6 hours *per week*). Their BMT is below 3 hours.

These subjects were volunteers who were informed and gave their full consent prior to participation (see Enclosed 16.1 below). They had to be unimpeded by neurological or musculoskeletal pathologies affecting running ability.

This sample didn't agree to statistics sample theory, because of the relevant novelty of the study (Colton, 1979; Blailar III et al., 1988; Matthews et al., 1988; Glantz, 1994; Venables, 2002).

Mean age (y), body mass (kg) and height (cm) of each group are presented in Table 16.1. One-way ANOVA for unrelated measures (and a post-*hoc* paired *t*-test, with Bonferroni correction) has shown only a significance in the factor 'age' if top runners have been compared to other runners ($p<0.001$). Yet, no other significances have been found in anthropometric dimensions.

TEST GROUP	Age (y) ± S.D.	Body Mass (kg) ± S.D.	Height (cm) ± S.D.
OCCASIONAL RUNNERS	33.1 ± 13.2	70.6 ± 3.4	175.9 ± 4.7
SKILLED RUNNERS	31.9 ± 11.8	67.3 ± 6.1	177.3 ± 4.0
TOP RUNNERS	42.6 ± 7.4	68.2 ± 4.9	177.8 ± 4.4

Table 16.1. Mean values of age, body mass and height.

Moreover, mean measured leg lengths (derived from the static trial in kinematic data; see par. 2 below) are illustrated in Table 16.2a (all lower limb: from Greater Trochanter to Lateral Malleolus) and Table 16.2b (just Thigh segment: from Greater Trochanter to Knee; and just Shank segment: from Knee to Lateral Malleolus), respectively. One-way ANOVA for unrelated measures (and a post-*hoc* paired *t*-test, with Bonferroni correction) has shown no significance in all leg measurements. As shown in literature, for humans, lower leg length is about 0.54*(hip height) (Aerts et al., 2000; Winter, 2005).

TEST GROUP	RIGHT LEG LENGTH (cm) ± S.D.	LEFT LEG LENGTH (cm) ± S.D.
OCCASIONAL RUNNERS	83.08 ± 3.57	82.85 ± 3.71
SKILLED RUNNERS	84.02 ± 4.12	83.04 ± 3.69
TOP RUNNERS	85.77 ± 6.30	84.80 ± 7.24

Table 16.2a. Mean values of leg length.

TEST GROUP	RIGHT THIGH LENGTH (cm) ± S.D.	LEFT THIGH LENGTH (cm) ± S.D.	RIGHT SHANK LENGTH (cm) ± S.D.	LEFT SHANK LENGTH (cm) ± S.D.
OCCASIONAL RUNNERS	42.16 ± 2.03	43.00 ± 2.80	40.93 ± 2.73	39.85 ± 2.26
SKILLED RUNNERS	44.47 ± 3.41	43.55 ± 2.65	39.55 ± 2.52	39.49 ± 2.22
TOP RUNNERS	45.09 ± 5.76	40.67 ± 2.93	44.36 ± 5.59	40.44 ± 3.71

Table 16.2b. Mean values of thigh and shank length.

Single subject anthropometric data are contained in the enclosed CD (Second Study, Chapter 16, Anthropometry and leg length).

2. PROTOCOL TEST

First of all, markers were placed on the anatomical landmark points. K4b^2 was arranged on the subject (Figure 16.1; see also chapter 4, par. 3.4.5 and par. 4.3).

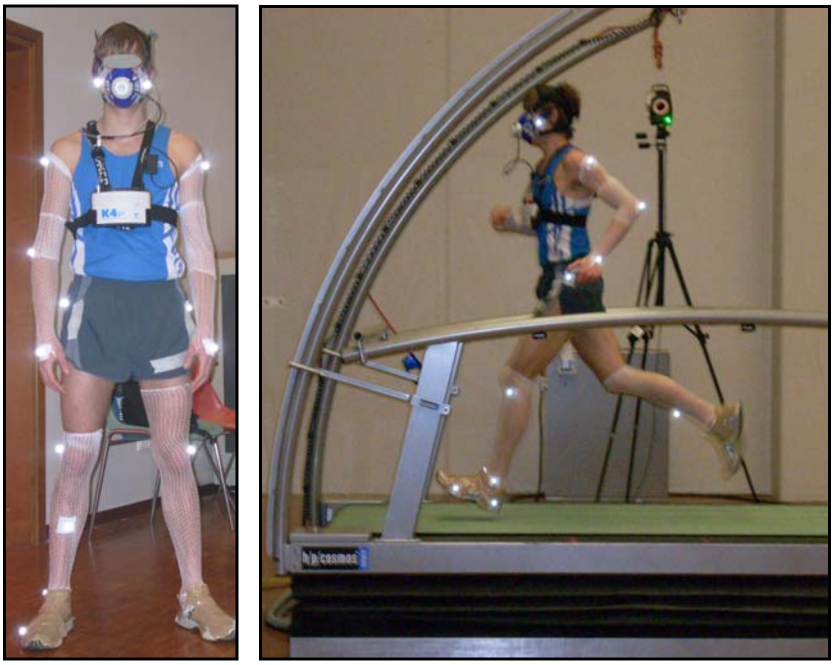
Figure 16.1. Markers and K4b^2 placed on a habitual runner (frontal view and lateral left one).

Secondly, each subject had to perform a single test session, including:

1. a brief period of familiarisation on the treadmill (at least 10 minutes, according to the documentation data; Bayat et al., 2005; Lavcanska et al., 2005; Teunissen et al., 2007);
2. a Magnetic Resonance Imaging (MRI), to record static anatomical symmetries (Figure 16.2; Siebenz et al., 1993; Mousavi et al., 2009);

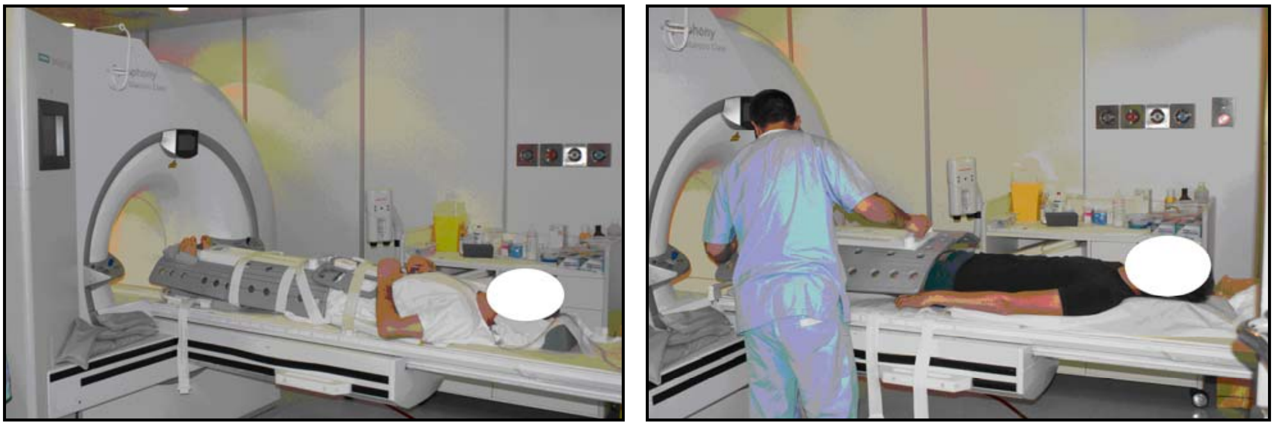
Figure 16.2. The instrumentation to acquire a MRI, in Borgo Roma Hospital (VR).

3. level running (Davies, 1980; Meyer et al., 2003) at 6 different speeds: from 2.22 m/s (= 8 km/h) to 5.00 m/s (= 18 km/h); step 0.56 m/s (= 2 km/h). Level running was performed in order to record both kinematic functional symmetries (Bhambhani et al., 1985; Novacheck, 1995; Weyand et al., 2000; Fletcher et al., 2009) and running economy (Steudel-Numbers et al., 2009). To sum up, each subject carried out at the most 6 trials.

On average, the test running session took 3 hours; and the MRI lasted 30 minutes.

Only five skilled runners and five top runners were able to complete all the running protocol (up to 5.00 m/s). Other subjects stopped at the speed of 4.44 m/s.

During each test, the subject had to run as naturally and regularly as possible (see also chapter 5, par. 2.1). He also had to keep to in the middle of the treadmill, looking straight ahead.

The 20 reflective markers were placed on the anatomical landmark points. The $K4b^2$ was adjusted to the subject (Zamparo et al., 2000).

After these preliminaries, 6 minutes of basic routine was proposed (Minetti et al., 2001; Donelan et al., 2002; Doke et al., 2004; Ardigò et al., 2005; Gottschall et al., 2005; Modica et al., 2005; Ortega et al., 2007; Bar-Haim et al., 2008; Sawicki et al., 2008; Zamparo et al., 2008; Mahaudens et al., 2009). The subject had to remain in a natural upright posture (Lejeune et al., 1998; Mahaudens et al., 2009; Plasschaert et al., 2009). This phase is fundamental to calculate the energy cost (see also chapter 17, par. 3).

Then, protocol running test started: each speed was maintained for at least 5 minutes. Indeed, this represents a time long enough to record an acceptable number of gait strides and corresponding physiological variables (Minetti et al., 1992; Jones et al., 1996; McArdle et al., 2001; Bramble et al., 2004; Gottschall et al., 2005; Modica et al., 2005; Chen et al., 2009).

A progressive incremental order was respected. Among the speeds, a rest period of at least 5 minutes was proposed. Before continuing the test, we had to control:

1. the heart rate, HR (e.g. before starting, HR is never higher than 100 bpm, according to McArdle et al., 2001; Willmore, 2004; Mian et al., 2006);
2. the respiratory exchange ratio, RER (e.g. before starting, RER is never higher than 1, according to Lejeune et al., 1998; McArdle et al., 2001; Willmore, 2004; Ardigò et al., 2005; Mian et al., 2006; Ortega et al., 2007; Sawicki et al., 2008; Mahaudens et al., 2009);
3. the individual's predisposition to continue and conclude the test.

If only one of these conditions failed, the test was stopped.

For two runners (1 occasional and 1 skilled runner), only one kinematic registration was performed at each speed. However, for all the others, during each run, three consecutive kinematic registrations were carried out: **a)** one at the end of the first minute (from 0.30 seconds to 1.00 minute); **b)** one in the middle of the test (from 2.30 to 3.30 minute); and **c)** one at the end of the last minute (from 4.30 to 5.00 minute). Their average value was used in further analysis (see chapter 18 and 20). To sum up, 400 trials/conditions were examined, in total.

As previously presented (see also chapter 5, par. 2.1), in order to avoid external influences on individual patterns of walking and running, subjects were never aware when each registration data began and/or stopped. Physiological parameters were continuously recorded by means of the telemetry portable metabograph K4b^2 (Hausswirth et al., 1997; Zamparo et al., 2000; Minetti et al., 2001; Ardigò et al., 2005; Zamparo et al., 2008).

All running testing was carried out utilising the Physiological Laboratory and the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University.

All MRI was carried out in the general Hospital of Borgo Roma in Verona with the kind collaboration of Dr. Faccioli and his staff.

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Department of Neurological and Visual Sciences
PhD Program in Exercise and Human Movement Science

THE RELATIONSHIP BETWEEN BODILY SYMMETRIES AND RUNNING ECONOMY IN HUMANS

Thank you for your taking part in this scientific experiment. Before starting, you will be given some information about why the exam is being carried out.

The main aim of this project is to verify both static anatomical and kinematic functional symmetries as important and relevant determinants of running economy.

To reach this goal, you will be expected to run on a treadmill, on level conditions, at 6 different incremental speeds (from 2.22 to 5.00 m/s; step 0.56 m/s).

20 reflective markers will be placed on the anatomical landmark points. A motion capture system will record kinematic data, in order to specify kinematic anatomical symmetries. At the same time, running economy will be recorded with the portable metabolic system K4b^2 (Cosmed). To analyse static anatomical symmetries, a MRI (Magnetic Resonance Imaging) will be carried out.

All running testing will be performed utilising the Biomechanics Laboratory of the Faculty of Exercise and Sport Science at Verona University. All MRI will be carried out in the general hospital of Borgo Roma in Verona.

We assure you that all data will remain anonymous and privacy will be guaranteed. Furthermore, data will only be utilized as regard this scientific research project.

Verona, date ………………

Tester’s signature

………………………………………………………………

Researcher’s signature

………………………………………………………………

Enclosed 16.1. Informed consent to participate in the second study.

Dipartimento di Scienze Neurologiche e della Visione
Corso di Dottorato in Scienze dell'Esercizio Fisico e del Movimento Umano

LA POSSIBILE RELAZIONE TRA SIMMETRIE CORPORE ED ECONOMIA DELLA CORSA NEI SOGGETTI UMANI

RingraziandoLa per avere aderito a questa sperimentazione scientifica, La informiamo sulla natura delle valutazioni che effettueremo e relative motivazioni.

Questo studio prevede di verificare sia simmetrie statiche anatomiche corporee che cinematiche funzionali come ipotetici predittori di un'importante indicatore di prestazione della corsa e cioè della sua economia.

Per potere raggiungere questo obiettivo, abbiamo la necessità di effettuare su di Lei prove di corsa in piano su ergometro trasportatore incrementandoLe progressivamente la 'velocità' (da 2.22 a 5.00 m/s; con uno step di 0.56 m/s).

Le saranno applicati, in maniera non invasiva e su principali punti di repere anatomici, marker riflettenti, i movimenti dei quali verranno registrati automaticamente e digitalmente da un sistema optoelettronico di analisi del movimento al fine di caratterizzare le simmetrie anatomiche cinematiche. Contemporaneamente, l'economia della corsa sarà registrata attraverso il metabolimetro portatile K4b^2 (Cosmed). Infine, attraverso una Risonanza Magnetica per Immagini (RMI), saranno registrate anche le simmetrie statiche anatomiche.

Le prove sperimentali verranno interamente effettuate presso il Laboratorio di Biomeccanica della Facoltà di Scienze Motorie. Le ricordiamo che tutti i dati raccolti sono strettamente coperti da privacy, utilizzati solo a scopo di ricerca scientifica e coperti da anonimato.

Verona, data ………………

Firma, per presa visione, del soggetto sottoposto alle valutazioni
……………………………………………………………………
Firma del ricercatore
……………………………………………………………………

Allegato 16.1. Consenso informato per partecipare allo studio.

Chapter 17

METHODS TO ANALYSE BODILY SYMMETRIES AND RUNNING ECONOMY

1. STATIC SYMMETRIES: MAGNETIC RESONANCE IMAGING

1.1. Introduction

To analyse MRI frontal sections, two different parallel approaches were developed.

In the first approach (**two dimensional analysis**: par. 1.2), MRI was analysed by means of two custom-written softwares: a) one commercial software, DicomWorks, version 1.3.5 (XP, UK), and b) one of the most utilized and widespread software, ImageJ, version 1.4.1 (XP, UK).

In the second approach (**three-dimensional analysis**: par. 1.3), a computerized image recognition (in LabVIEW, version 8.6) has been developed.

In the following sections, we will therefore focus on these two methods.

1.2. The two-dimensional (2D) analysis

1.2.1. Introduction

Particularly, at T1 relaxation (see also chapter 15, par. 2.4.4), bone and lipid contents are visualised with colour white (in this case, there is a lot of medulla inside), whereas muscle content is visualised with colour black (there is a lot of water inside). However, at T2 relaxation, structures containing water or similarity are represented with colour white.

Most of our MRI studies are T1 relaxation types (Figure 17.1).

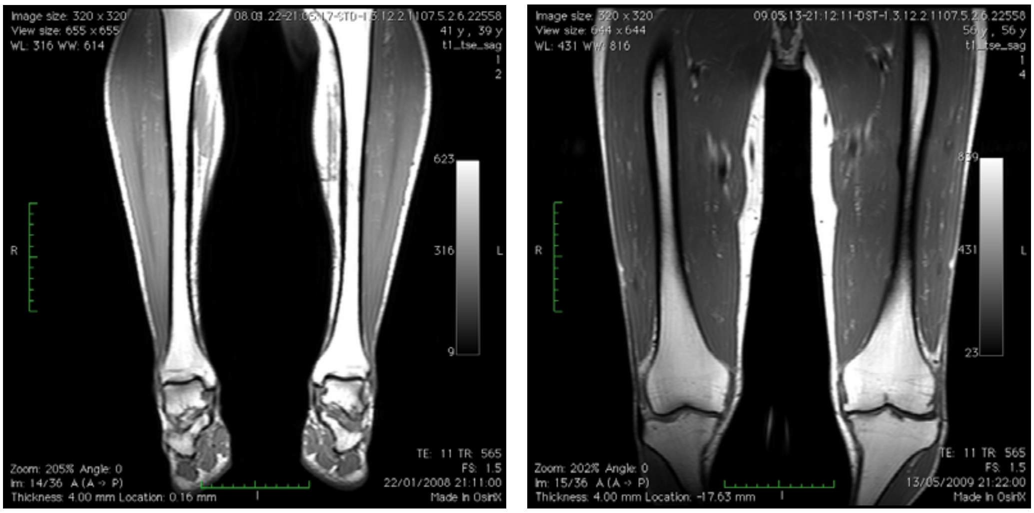
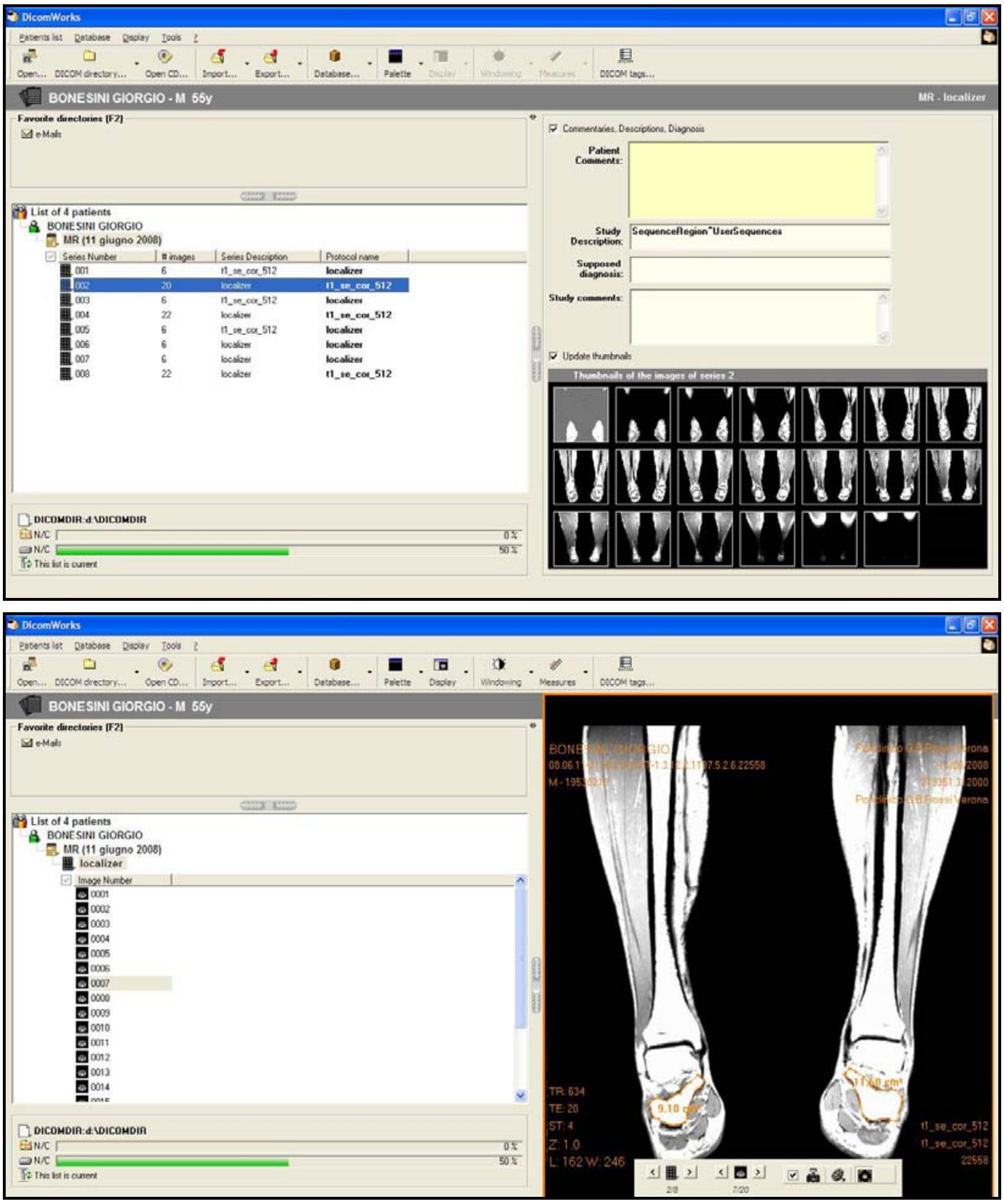


Figure 17.1. MRI T1 relaxation types: leg region (on the left) and knee joint (on the right).

In our study, each anatomical region and each subject give us from 20 to 36 images or sections that could be analysed. As supposed, these sections vary among all subjects.

1.2.2. Custom-written software

Dicom Works software (Figure 17.2) allowed us to isolate each patient/subject individuating the main regions of the lower limb (foot and shank area, knee and thigh area) and pelvis (including hip joint), in terms of areas, volumes and angles. In such a way, all anatomical regions were visualized and then saved (modalities 'export' and 'save a picture file') in order to be studied with the software ImageJ.

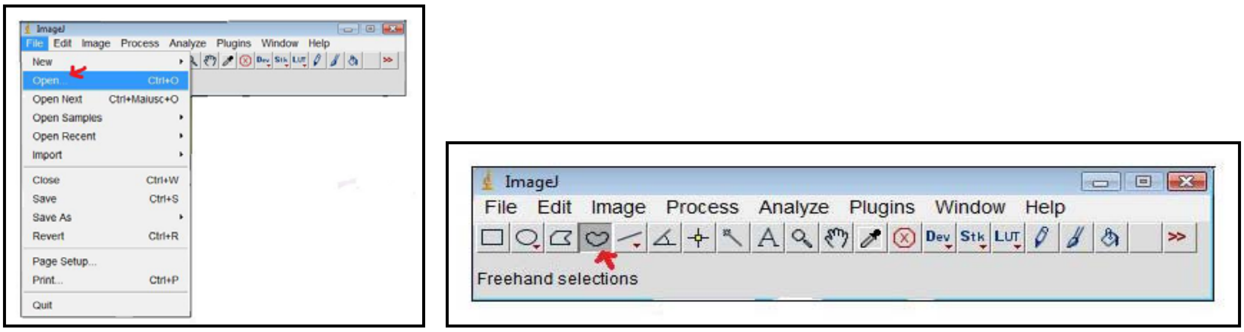


Series Number	# images	Series Description	Protocol name
001	6	t1_se_cor_512	localizer
002	20	localizer	t1_se_cor_512
003	6	t1_se_cor_512	localizer
004	22	localizer	t1_se_cor_512
005	6	t1_se_cor_512	localizer
006	6	localizer	localizer
007	6	localizer	localizer
008	22	localizer	t1_se_cor_512

Study Description: SequenceRegion^UserSequences

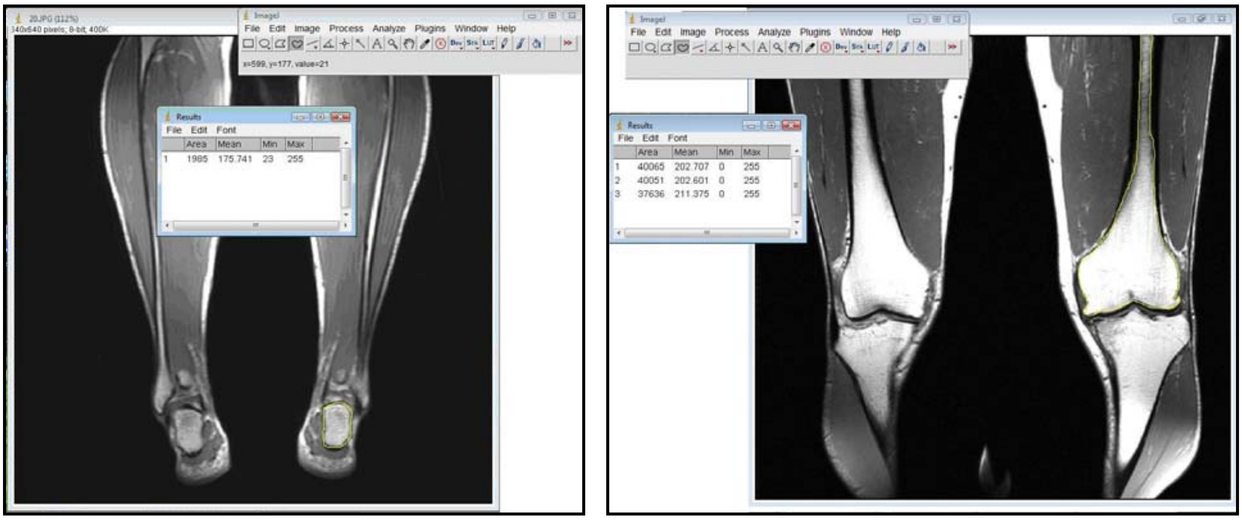
Figure 17.2. Dicom Works 1.3.5 software.

ImageJ software (Figure 17.3a and 17.3b) allowed us to measure these main regions of the lower limb and pelvis, by using suitable pallets and the movement of the mouse.



Freehand selections

Figure 17.3a. Screen captures of the ImageJ 1.4.1 software.

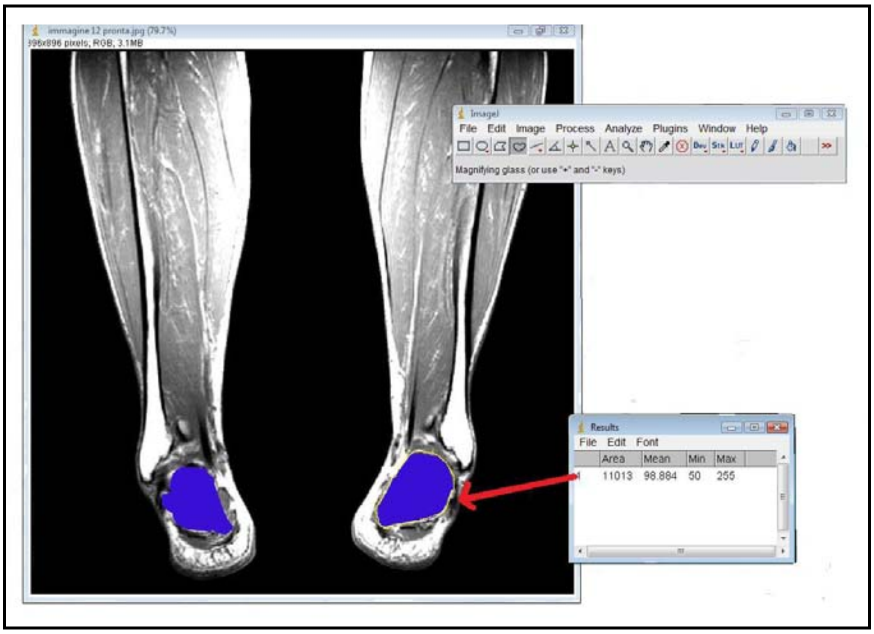


```
x=599, y=177, value=21
```

```
        Area    Mean     Min   Max
1       1985    175.741  23    255
```

```
        Area    Mean     Min   Max
1       40065   202.707  0     255
2       40051   202.601  0     255
3       37636   211.375  0     255
```

Figure 17.3b. ImageJ software: foot area (on the left) and knee region (on the right).



Magnifying glass (or use "+" and "-" keys)

```
        Area    Mean     Min   Max
1       11013   98.884   50    255
```

Figure 17.3c. ImageJ software and its pallets (foot segment: heel area).

Because its large possibility of measuring and obtaining information on the anatomical parameters, we decided to use this last software. By individuating specific regions and by using suitable pallets (Figure 17.3c), it makes possible to quantify the number of pixels of both right and left sides. All these measures were made by a same operator (according to Manning et al., 1994).

In order to respect the inter-reliability, each character was measured three times for a subject (Manning et al., 1994). An example is illustrated in Table 17.1.

SUBJECT (#)	SECTION (nr.)	HEEL AREA	RIGHT SIDE	LEFT SIDE
OR #6	**11**	First Measurement	1248	1025
		Second Measurement	1128	1114
		Third Measurement	**1260**	**1117**
SR #1	**14**	First Measurement	**1066**	**1220**
		Second Measurement	1125	1254
		Third Measurement	1051	1192
TR #1	**16**	First Measurement	**1837**	**1748**
		Second Measurement	1760	1749
		Third Measurement	1721	1656

Table 17.1. Example of inter-reliability criterion.

Therefore, we have decided to take the best measurement (Table 17.1, in bold) as the referring value. In effect, the best visible and graphical outlines in each testing image constitute the operator's criterion in selecting such best measurement.

1.2.3. Foot segment and shank area

Therefore, in occasional, skilled and top runners' images, the same operator measured:

a) the <u>heel area</u>, in the first section showing the tibiae (Figure 17.4);

Figure 17.4. Heel area showing the tibiae in an individual OR (rear view).

b) the <u>same heel area</u>, in the first section showing the fibulae (Figure 17.5a) and, for comparison, the adjacent section with un-shown fibulae (Figure 17.5b);

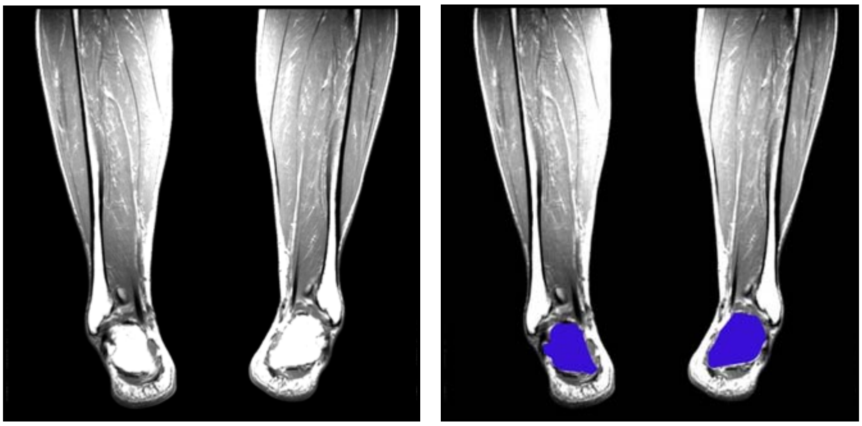


Figure 17.5a. Heel area showing the fibulae in an individual OR (rear view).

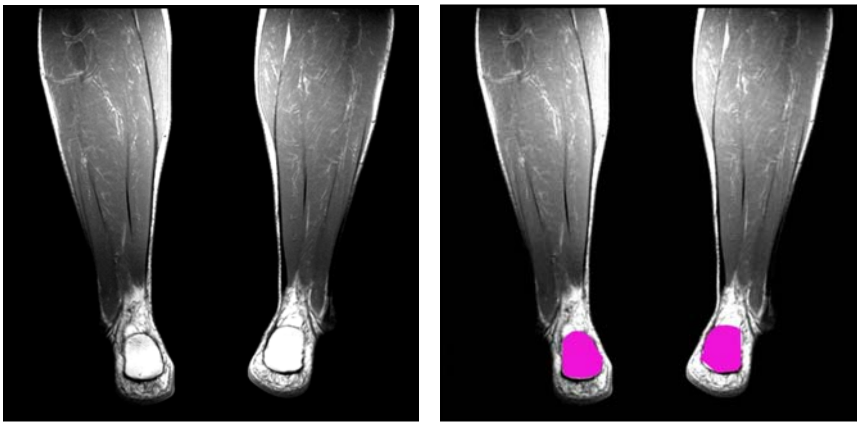


Figure 17.5b. Heel area un-showing the fibulae in an individual OR (rear view).

c) the major heel area (Figure 17.6);

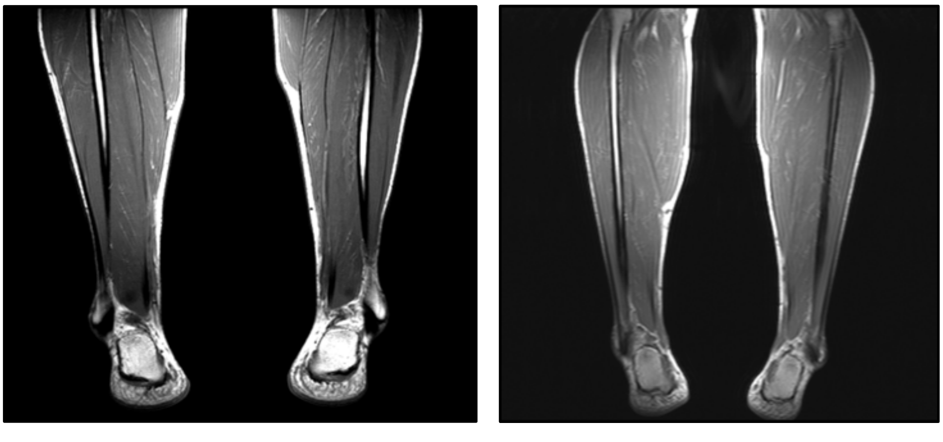


Figure 17.6. The major heel area (rear view) in an individual OR (on the left) and in an individual SR (on the right).

d) the heel bone area, in the first section where it is shown (Figure 17.7);

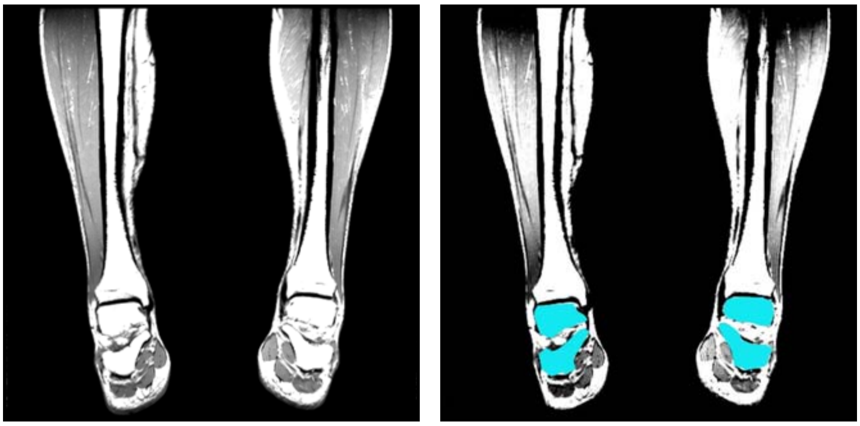


Figure 17.7. Heel bone area in an individual OR (rear view).

e) the fibula area, in the first section where it is shown;

f) the total area of the shank, in the first section showing the tibiae (Figure 17.8);

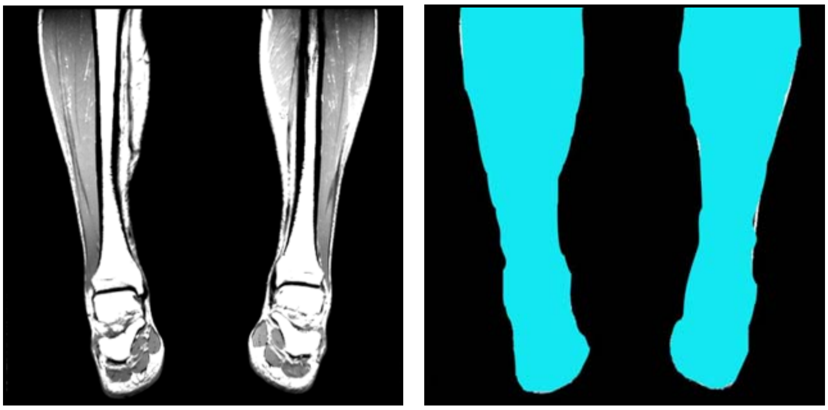


Figure 17.8. Shank total area (muscles and bones) in an individual OR (rear view).

g) the area of shank muscles, in the first section showing the tibiae (Figure 17.9);

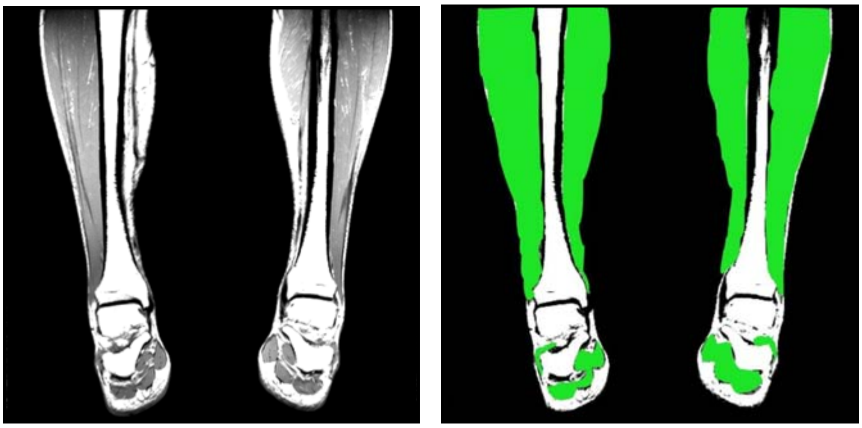


Figure 17.9. Shank muscle area in an individual OR (rear view).

h) the percentage of shank muscles, in the first section showing the tibiae. This value is obtained as:

$$\%\text{shankmuscles} = \frac{\text{shankmusclearea}}{\text{shanktotalarea}} \quad [\text{Eq. } 17.1]$$

i) the <u>area of shank bones</u>, in the first section showing the tibiae (Figure 17.10);

Figure 17.10. Shank bone area in an individual OR (rear view).

j) the <u>percentage of shank bones</u>, in the first section showing the tibiae. This value is obtained as:

$$\%\text{shankbones} = \frac{\text{shankbonearea}}{\text{shanktotalarea}} \quad [\text{Eq. } 17.2]$$

k) <u>the percentage of other shank tissues</u>, in the first section showing the tibiae. This value is obtained as:

$$\%\text{shanktissues} = \text{shanktotalarea} - (\text{shankmusclearea} + \text{shankbonearea}) \quad [\text{Eq. } 17.3]$$

In order to simplify and better summarize the results, specific signs have been used:

FOOT SEGMENT AND SHANK AREA	SIGN
Heel area, in the first section showing the tibiae	**L1**
Heel area, in the first section showing the fibulae	**L2**
Heel area, in the first section un-showing the fibulae	**L3**
Heel bone area, in the first section showing the tibiae	**L4**
Total area of the shank, in the first section showing the tibiae	**L5**
Area of shank muscles, in the first section showing the tibiae	**L6**
Area of shank bones, in the first section showing the tibiae	**L7**
Percentage of shank muscles, in the first section showing the tibiae	**%L8**
Percentage of shank bones, in the first section showing the tibiae	**%L9**
Percentage of other shank tissues, in the first section showing the tibiae	**%L10**

Table 17.2. Signs of foot segment and shank area.

As a consequence, in chapter 18 results will be illustrated and discussed using these proper signs (Table 17.2).

1.2.4. Knee joint and thigh area

In occasional, skilled and top runners' images, the same operator measured:

a) the femur area, in the first section showing the thigh bones (femur bones) (Figure 17.11);

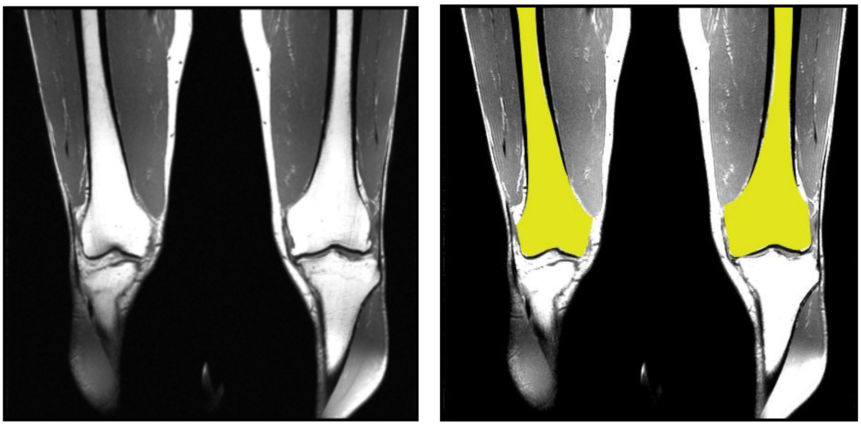


Figure 17.11. Femur area showing the thigh bones in an individual OR (rear view).

b) the femur area, in the first section showing the thigh bones and the tibiae (Figure 17.12);

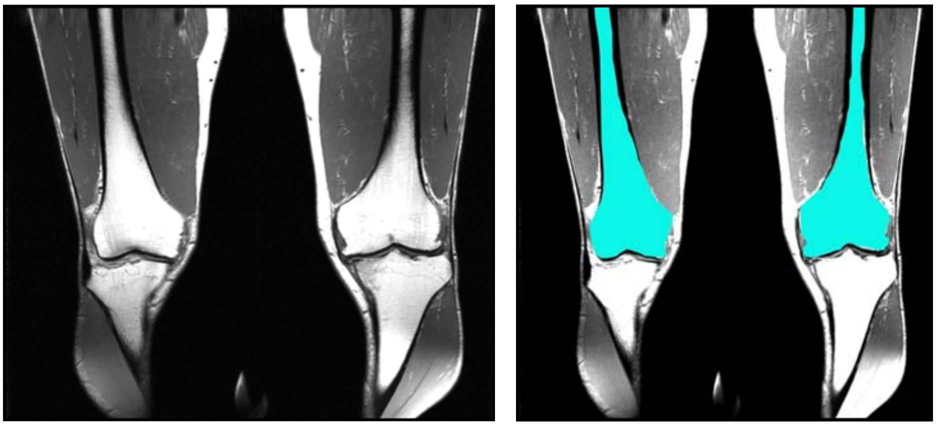


Figure 17.12. Femur area showing the thigh bones and the tibiae in an individual OR (rear view).

c) the total area of the thigh, in the first section showing the thigh bones (Figure 17.13);

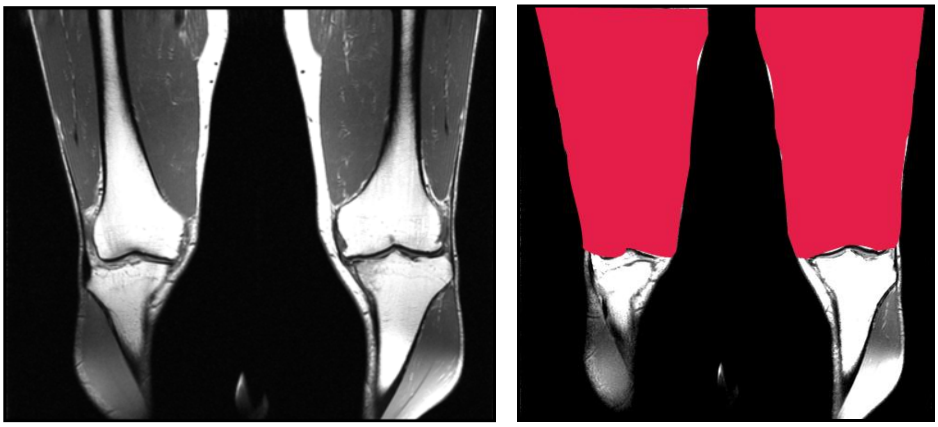


Figure 17.13. Thigh total area (muscles and bones) in an individual OR (rear view).

d) the area of thigh muscles, in the first section showing the thigh bones (Figure 17.14);

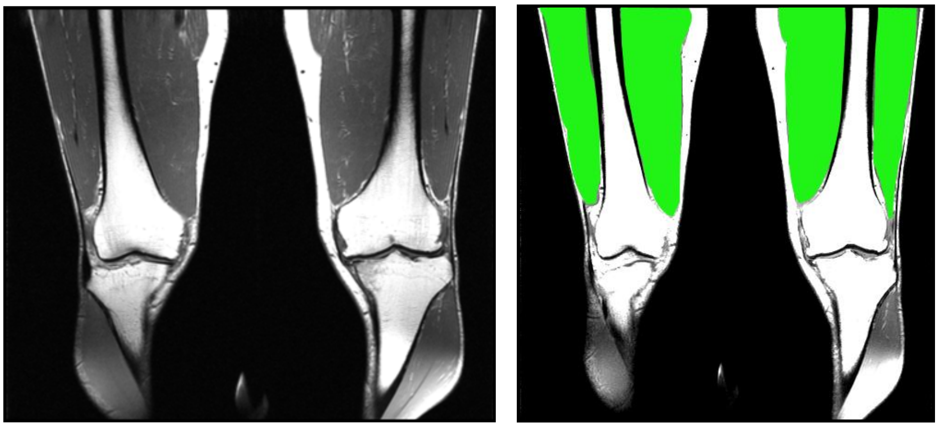


Figure 17.14. Thigh muscle area in an individual OR (rear view).

e) the percentage of thigh muscles, in the first section showing the thigh bones. This value is obtained using Equation [17.1] and putting in it the value for thigh instead of shank;
f) the area of thigh bones, in the first section showing the thigh bones (Figure 17.11 above);
g) the percentage of thigh bones, in the first section showing the thigh bones. This value is obtained using Equation [17.2] and putting in it the value for thigh instead of shank;
h) the percentage of other thigh tissues, in the first section showing the thigh bones. This value is obtained using Equation [17.3] and putting in it the value for thigh instead of shank.

In order to simplify and better summarize the results, specific signs have been used:

KNEE JOINT AND THIGH AREA	SIGN
Femur area, in the first section showing the femur	**T1**
Femur area, in the first section showing the femur and the tibiae	**T2**
Total area of the thigh, in the first section showing the femur	**T3**
Area of thigh muscles, in the first section showing the femur	**T4**
Area of thigh bones, in the first section showing the femur	**T5**
Percentage of thigh muscles, in the first section showing the femur	**%T6**
Percentage of thigh bones, in the first section showing the femur	**%T7**
Percentage of other thigh tissues, in the first section showing the femur	**%T8**

Table 17.3. Signs of knee joint and thigh area.

As a consequence, in chapter 18 results will be illustrated and discussed using these proper signs, too (Table 17.3).

1.2.5. Hip joint and pelvic area

Because of the low resolution of single images, it was very difficult to isolate (and measure) number of pixels in the hip joint and the pelvic region. Indeed, anatomical limits of femur and pelvis were not so clear, marked and distinct. Therefore, we decided not to involve this area in our bi-dimensional analysis.

Hopefully, future MRI protocols will better focus on such a region in order a) to easier analyse its complex architecture, b) to cut off its main segments and sectors, and c) to wholly complete our analysis.

1.2.6. Mathematical analysis

Firstly, the number of pixels in each area was measured both for the left and the right side. As previously reported (see par. 1.2.2 above), all these measures were made for three times by a same operator.

Secondly, for occasional, skilled and top runners, we calculated:

a) for each subject, the mathematical ratio/connection (the so-called *Static Symmetry Ratio*, SSR) between the average value in both left and right side (Manning et al., 1994), as:

$$\mathrm{SSR} = \frac{\mathrm{LeftArea\ \ (squarepixel)}}{\mathrm{RightArea\ \ (squarepixel)}} \text{ [Eq. 17.4]}$$

b) for each running group, beginning from the values obtained at the point a), the mean value and the standard deviation of these relations (mean ± standard deviation (S.D.));

c) for each running group, the corresponding coefficient variation (CV), as:

$$CV = \left(\frac{S.D.}{Mean}\right) \bullet 100 \text{ [Eq. 17.5]}$$

In this way, it became possible to compare:

- the left to the right side in a same subject. Specifically, a SSR = 1 means that there are no differences among the sides; however, a SSR < 1 means that the pixel number in the right side is greater than in the left; on the opposite side, a SSR > 1 means that the pixel number in the left side is greater than in the right;
- subjects of the same running ability to subjects of different running ability.

1.3. The three-dimensional (3D) analysis

1.3.1. Introduction

In the second approach, a biomedical engineer who collaborates with prof. A.E. Minetti (Milan University) has developed a computerized image recognition algorithm.

This program *ad hoc* has been written in LabVIEW, a software for data processing working both on a Windows-based PC and on a Macintosh notebook (version 8.6), making thus possible:

- firstly, to open the Dicom files (see par. 1.2.2 above) coming from the MRI;
- secondly, to decide both which subject/patient will be analysed and which anatomical region will be selected and investigated (i.e. foot segment and shank area, knee joint and thigh area and pelvic region);
- lastly, to mathematically three-dimensional analyse each human region/district.

In the following sections, the main steps of this approach will be illustrated and summarized.

1.3.2. Steps of the computerized image analysis

A. Once the patient and his anatomical region have been chosen, the program starts to load and open the sections that characterize that district. In detail, every section is a bi-dimensional image and the used protocol provided 36 section for one district each one made of 320 *per* 320 pixel.

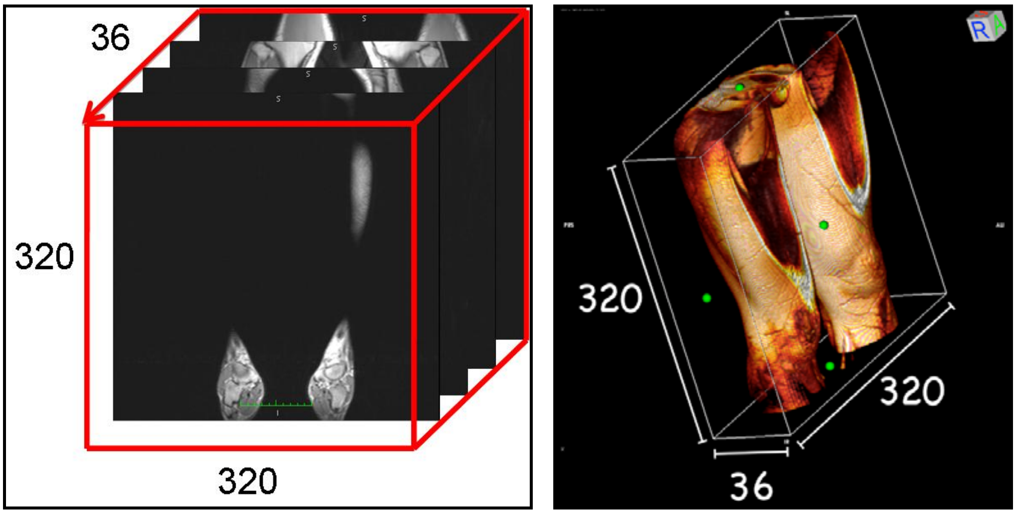
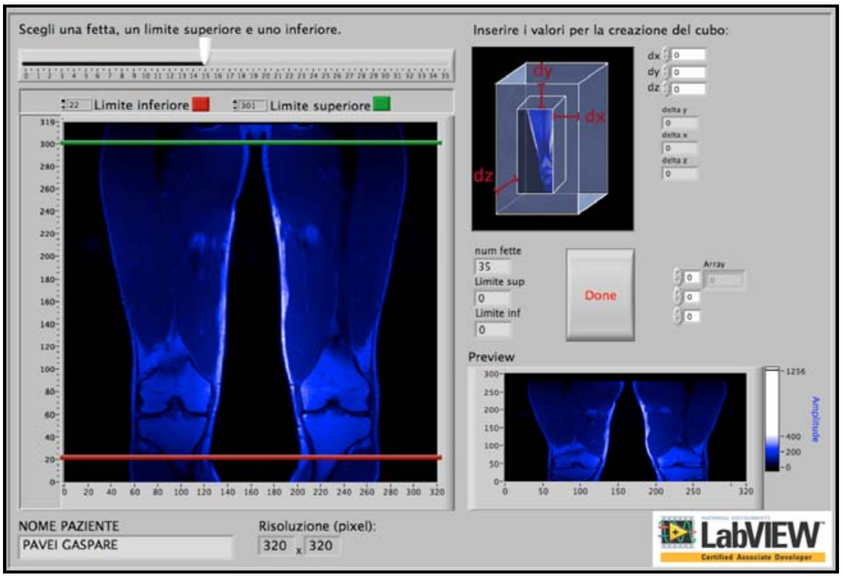


Figure 17.15. The mathematical matrix in three dimensions (320·320·36; courtesy of Elena Seminati).

Every pixel is a numeric value that identifies the image grey level: black is 0 and the more the value increases, the more the colour is close to the white. Hence, the software builds a numeric matrix characterised by three-dimensions (Figure 17.15): 1) 320 pixel for the width; 2) 320 pixel for the height; and 3) 36 sections.

B. Once the sections have been loaded, it becomes possible to choose and select the limits (green for the superior limit and red for the inferior limit) of that area/zone in which the comparison would be applied (Figure 17.16). Clearly, as illustrated in the Preview (see Figure 17.16 and 17.20 below; for example, 160 pixel), the height dimensions will be reduced in the matrix: from 320 to the value chosen for the maximum height.


Scegli una fetta, un limite superiore e uno inferiore.

22 Limite inferiore 🟥

301 Limite superiore 🟩

Inserire i valori per la creazione del cubo:

dx 0
dy 0
dz 0
delta y 0
delta x 0
delta z 0

num fette 35
Limite sup 0
Limite inf 0

Array 0 0 0

NOME PAZIENTE PAVEI GASPARE

Risoluzione (pixel): 320 x 320

Figure 17.16. Selected superior and inferior limits in the foot segment and shank area (courtesy of Elena Seminati).

C. Chosen image will be divided in two equal parts. In such a way, the right portion could be compared to the left one and *vice versa* (Figure 17.17).

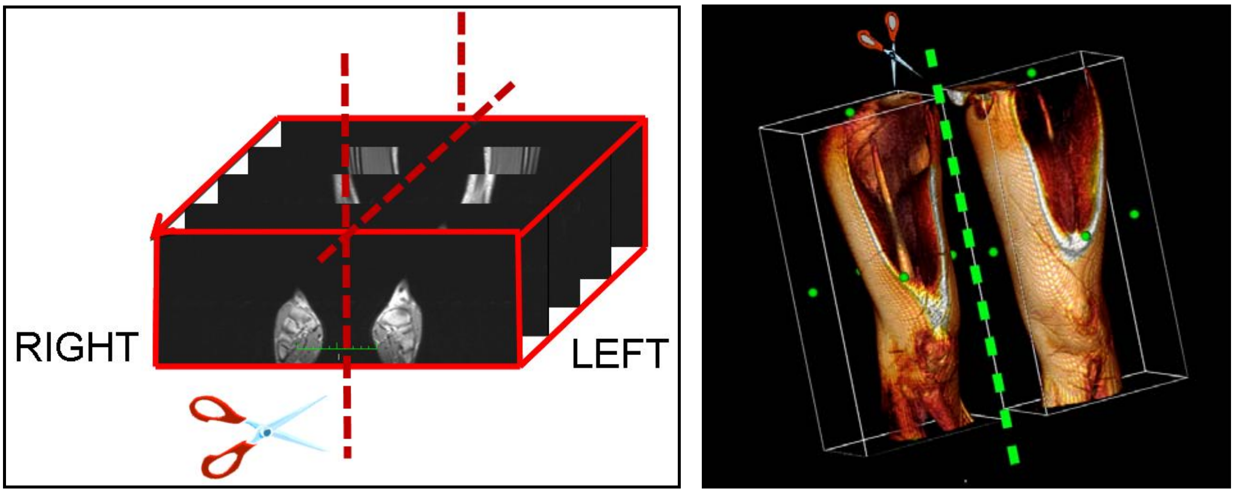


Figure 17.17. The right and the left portions/regions (courtesy of Elena Seminati).

D. In this step, the program is able to mirror the left region in(to) the right, obtaining two different images (Figure 17.18).

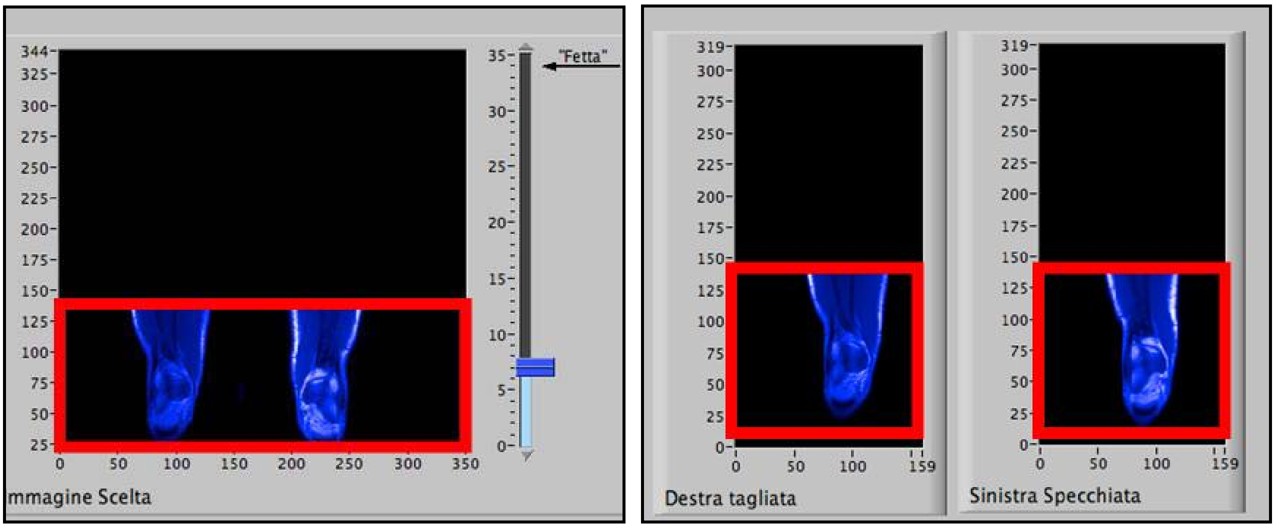


Figure 17.18. The left region is mirrored into the right region (courtesy of Elena Seminati).

E. After duplicating the image specularly, the program superimposes the 3D image of the left region (the mirrored image) on the 3D image of the right.

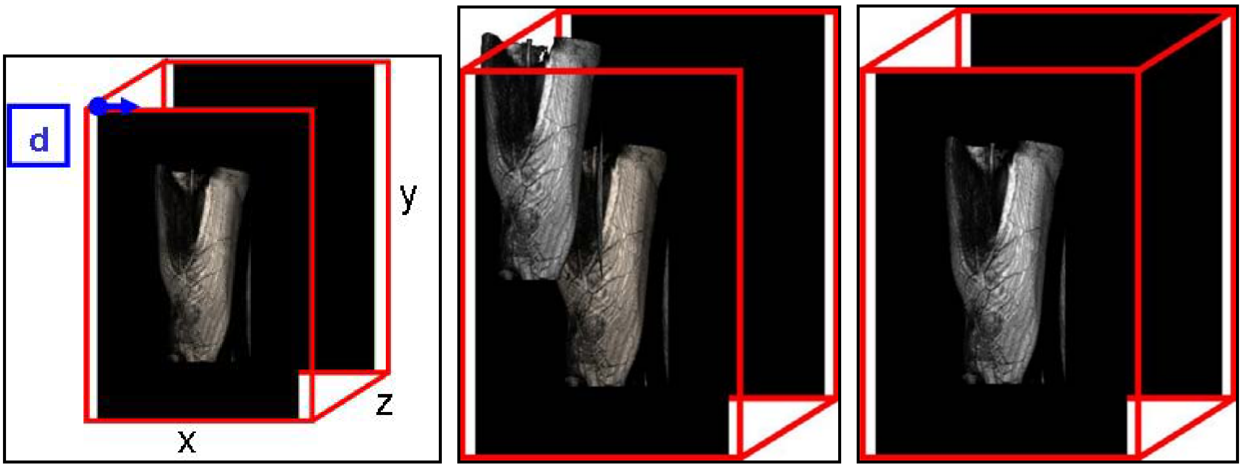


Figure 17.19. How the left region tries to superimpose the right region (courtesy of Elena Seminati).

As shown in Figure 17.19, the left region moves of a distance of *d* among the three different dimensions x, y and z (respectively, right and left; up and down; forward and backward) to find the best position in order to look as much as possible like the right one. This best position ensures the lowest differences between right region and left mirrored region and therefore the best superimposition between these two regions.

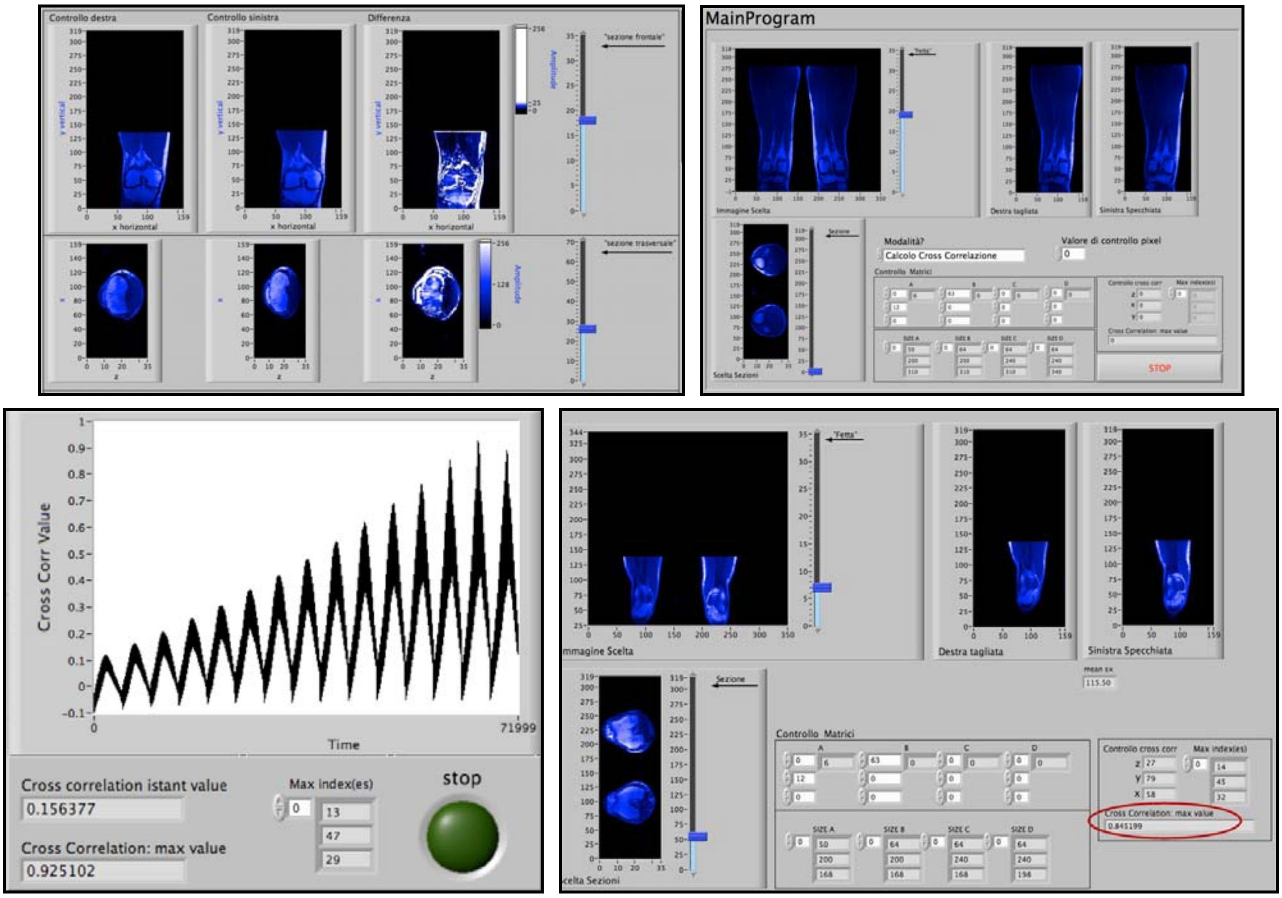
F. Mathematically, the algorithm applies to each numeric value (called *voxel*), corresponded to a specific grey intensity for the right region and for the left region (see point A above), a mathematical formula that allows to compute a cross-correlation coefficient *r*:

$$r = \frac{\Sigma[(p_r(i) - mx_r) \bullet (p_l(i - \Delta) - mp_l)]}{\sqrt{\Sigma(p_r(i) - mp_r)^2} \bullet \sqrt{\Sigma(p_l(i - \Delta) - mp_l)^2}} \quad \text{[Eq. 17.6]}$$

This formula is an example of what happens in two dimensions:

- p_r represents the right position;
- p_l the left position;
- *m* the average value of x or y coordinates;
- *i* the number of pixel of the images along each dimension;
- Δ is the displacement of a region in respect to the other.

The resulting screen captures are represented (step by step) in Figure 17.20: in the red circle, the cross-correlation coefficient has been highlighted.



Modalità? Calcolo Cross Correlazione

Valore di controllo pixel 0

Cross correlation istant value 0.156377

Cross Correlation: max value 0.925102

Max index(es) 0 13 47 29

mean sx 115.50

Controllo cross corr z 27 y 79 x 58

Max index(es) 0 14 45 32

Cross Correlation: max value 0.845199

Figure 17.20. Screen captures of three-dimensional analysis software (in LabVIEW) (courtesy of Elena Seminati).

In such a way, a cross-correlation coefficient has been obtained. Clearly, it ranges from 0 to 1 and it constitutes an index of static symmetry. In fact, if *r* tends to(wards) 1, this means that right and left images are very similar; *vice versa*, if *r* is near to 0, this means that right and left images are very different. Finally, it is also possible to calculate the corresponding determination coefficient (r^2). Between this mathematical coefficient and the symmetry index (previously described in chapter 8), there is a high and close correspondence.

In order to state the reliability of this symmetry coefficient, three different trials were made: 1) in a same subject, the right leg (image) has been compared to the left leg (image). The resulting value of *r* was 1, suggesting that the two images are equals; 2) repeated measures of the same subject were carried out. The values of *r* differed very little (i.e. 0.911749 *versus* 0.915672); and 3) in two different subjects, the right leg of the first subject was compared to the left leg of the second subject. The lowest values of *r* were obtained.

2. DYNAMIC SYMMETRIES

From kinematic data, Digital Locomotory Signature and corresponding Symmetry Index were calculated (and graphically represented) for each runners group (Karamanidis et al., 2003). In order to obtain this dynamical information concerning the movement of the BCOM, the same approach widely presented (and discussed) in the first study (see chapter 6, 7 and 8) was developed.

Single files *.bcm and *.res format in all runners are contained in the enclosed CD (Second Study, Chapter 17, File *.BCM and file *.RES). Furthermore, single templates are in the sections Average template versus speed, Spreadsheet different groups and Template level running.

Average values of speed obtained with the *.vi *Motion Analysis Filter* (described in chapter 6, par. 2.1) and used in graphical representations (see chapter 18 onwards) are:

RUNNING SPEED	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
2.22 m/s (= 8 km/h)	2.19	2.23	2.24
2.78 m/s (= 10 km/h)	2.74	2.80	2.66
3.33 m/s (= 12 km/h)	3.25	3.34	3.33
3.89 m/s (= 14 km/h)	3.90	3.89	3.89
4.44 m/s (= 16 km/h)	4.45	4.45	4.44
5.00 m/s (= 18 km/h)	result no available	4.99	5.00

Table 17.4. Average speed values in all runners.

One-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) has highlighted slightly significant differences: a) in occasional runners: speeds of 2.22 m/s ($p<0.05$), 2.78 m/s ($p<0.05$) and 3.33 m/s ($p<0.01$); and b) in top runners: speed of 2.78 m/s ($p<0.01$). Single values in all runners are contained in the enclosed CD (Second Study, Chapter 17, Template level running: average values of speed).

In this way, we could investigate the pattern of dynamic symmetries as a function of both speed and running ability.

2.1. Discarded tests

It is important to note that, in addition to the exceptions illustrated in chapter 16 (par. 2), during the kinematic data analysis it became evident that other tests had to be rejected due to various and unexpected reasons. Consequently, some tests were discarded. Particularly:

- two kinematic recordings (from 2.30 to 3.30 and from 4.30 to 5.00 minute), in running at 3.33 m/s for 1 occasional runner;
- the kinematic recording, in running at 3.89 m/s for 1 occasional runner;
- two kinematic recordings (from 2.30 to 3.30 and from 4.30 to 5.00 minute), in running at 3.89 m/s for 1 occasional runner;
- one kinematic recording (from 4.30 to 5.00 minute), in running at 3.89 m/s for 1 occasional runner;
- the kinematic recording, in running at 4.44 m/s for 1 occasional runner;

- two kinematic recordings (from 2.30 to 3.30 and from 4.30 to 5.00 minute), in running at 4.44 m/s for 1 occasional runner;
- one kinematic recording (from 4.30 to 5.00 minute), in running at 4.44 m/s for 1 occasional runner;
- one kinematic recording (from 4.30 to 5.00 minute), in running at 4.44 m/s for 1 skilled runner;
- the kinematic recording, in running at 5.00 m/s for 1 skilled runner;
- one kinematic recording (from 4.30 to 5.00 minute), in running at 5.00 m/s for 1 skilled runner;
- one kinematic recording (from 0.30 to 1.00 minute), in running at 3.89 m/s for 1 top runner;
- one kinematic recording (from 0.30 to 1.00 minute), in running at 4.44 m/s for 2 top runners;
- one kinematic recording (from 0.30 to 1.00 minute), in running at 5.00 m/s for 1 top runner.

On the whole, **17 trials** were deleted.

3. METABOLIC COST: A RELEVANT DETERMINANT OF RUNNING ECONOMY

3.1. Introduction

In general, the cost of progression (metabolic or energy cost, C) is an important and relevant determinant of running economy. Indeed, each type of human locomotion is characterised by a specific metabolic cost (Margaria et al., 1938; 1963; Andolf et al., 1976; Mayhew, 1977; Dal Monte, 1983; di Prampero, 1985; 1986; Sparrow et al., 1987; Saibene, 1990; Steudel, 1990; Brueckner et al., 1991; Bourdin et al., 1993; Holt et al., 1995; di Prampero, 1997; Dalleau et al., 1998; Roberts et al., 1998; Hill, 1999; Sparrow, 2000; Wickler et al., 2000; Zamparo et al., 2000; Minetti et al., 2001; Weineck, 2001; Frost et al., 2002; Hamilton et al., 2002; Minetti et al., 2002; McCann et al., 2003; Steudel-Numbers, 2003; Minetti, 2004; Saunders et al., 2004; Ardigò et al., 2005; Beneke et al., 2005; Mian et al., 2006; Abe et al., 2007; Steudel-Numbers et al., 2007; Teunissen et al., 2007; Capelli et al., 2008; Ferretti et al., 2008; Richards, 2008; Zamparo et al., 2008; Houdijk et al., 2009; Peyrot et al., 2009).

The metabolic cost of locomotion has been linked to a number of anthropometric and kinematic characteristics that differ between men and women. These include factors such as: a) body size; b) body mass; c) distribution; d) stride length and/or stride frequency; and e) range of motion (Williams, 1985; Kang et al., 2002).

Specifically, the evaluation of the metabolic cost constitutes an additional characteristic during laboratory test that may help scientists a) to ascertain the effectiveness of the training procedure and

b) to evaluate the technique performed (Bunc et al., 1989). Importantly, metabolic cost represents 'a discriminating parameter in endurance performance' (Hill, 1999).

But '*how is it important to know this physiological variable?*' Or in other words, '*does a relationship between metabolic cost and running economy exist?*'

Therefore, it becomes necessary to focus on this central variable (metabolic cost) in order to better understand the behaviour of running economy in different testing conditions. Indeed, it is well known how an inverse relationship exists between these two parameters: indeed, running economy is the reciprocal of metabolic cost.

In the following sections, the main information and peculiarities of metabolic cost will be presented and discussed.

3.2. Definition of metabolic cost

Margaria (1938) was the pioneer. Indeed, in a classical study of the energy expenditure of walking and running at different speeds and on different gradients, he had introduced a similar parameter: e.g. the energy consumed *per* unit distance covered.

Before continuing, it becomes fundamental to remember the distinction between 'the cost *per* unit time' (cost of locomotion) and 'the cost *per* unit distance' (cost of transport), as suggested both in Mahaudens et al. (2009) and in Steudel-Numbers et al. (2009).

For a given subject, the metabolic cost C is the quotient of net metabolic power, divided by speed and it has the physical dimensions of force (Margaria, 1938; see Equation [17.7]).

Particularly, at sub-maximal speeds (Zamparo et al., 2000), it could be expressed as:

$$C = \frac{V'O_2 - V'O_2 rest}{s} \quad \text{[Eq. 17.7a] or}$$

$$C = \frac{V'O_2 \bullet ((4.94 \bullet RER) + 16.04) - V'O_2((4.94 \bullet RER) + 16.04)rest}{s} \quad \text{[Eq. 17.7b]}$$

where $V'O_2$-$V'O_2 rest$ is the net metabolic power (a measure of aerobic power, in ml/min·kg; di Prampero, 1986; Sparrow, 2000; Zamparo et al., 2000; McArdle et al., 2001; Minetti, 2004; Nakai et al., 2009); *RER* is the respiratory exchange ratio; and *s* is the average speed of progression (in m/s).

Moreover, to compare subjects of different sizes, the C is usually expressed as the quotient of net metabolic power divided by the product of speed times body weight, that has also the physical dimensions of power (Bergh et al., 1991; Cerretelli, 2001; McArdle et al., 2001; Davies et al., 2003; Saunders et al., 2004; Schwartz et al., 2006; Zakeri et al., 2006; Abe et al., 2007; Steudel-Numbers et al., 2007; Capelli et al., 2008; Zamparo et al., 2008; see Equation [17.8]). In this last case, it could be expressed as:

$$C = \frac{V'O_2 - V'O_2 rest}{s \bullet m} \text{ [Eq. 17.8a] or}$$

$$C = \frac{V'O_2 \bullet ((4.94 \bullet RER) + 16.04) - V'O_2((4.94 \bullet RER) + 16.04)rest}{s \bullet m} \text{ [Eq. 17.8b]}$$

where *V'O*$_2$*-V'O*$_2$*rest* is the net metabolic power (in ml/min·kg); *RER* is the respiratory exchange ratio; *s* in the average speed of progression (in m/s); and *m* is the body mass (in kg).

Furthermore, the metabolic cost of running, defined as the amount of energy consumed *per* unit of distance, reflects the sum of both aerobic and biomechanical demands (Slawinski et al., 2004).

In current literature, there are different theories concerning the determinants of metabolic cost: **1)** some researchers have demonstrated that primary biomechanical factors determine this variable in running. For instance, they could be: a) the magnitude; and b) the rate of muscular force generation to counteract the effect of gravity and to operate the spring-like properties of the muscle-tendon system (Kram, 2000; McArdle et al., 2001; Teunissen et al., 2007); and **2)** recently, it has been demonstrated that the metabolic cost has been shown to be correlated to vertical displacement and hence the potential energy changes of the BCOM (Houdijk et al., 2009).

Therefore, it becomes very difficult to distinguish between work and force/time contributions (Doke et al., 2004). Among the others, it could be hypothesised that a combination of factors influencing human locomotion (i.e. force, work, gravity, power and so on) properly determines the metabolic cost.

Finally, many factors are known or hypothesised to influence metabolic cost such as environmental conditions, participant specificity and metabolic modifications (i.e. training status, fatigue and so on (Hausswirth et al., 2001)).

3.3. The independence between metabolic cost of running and speed/gradient

Differently to what happens in human walking (Figure 17.21, left graph; Ralston, 1958; Cotes et al., 1960; Sparrow, 2000; Anderson et al., 2001) and in other sports (such as skiing, country-cross skiing and skating: Figure 17.21, left graph; di Prampero, 1985; Saibene et al., 2003), because of the linear relationship between oxygen consumption (V'O$_2$) and running speed, the total energy requirement for running at a given distance (steady rate) is about the same regardless of speed (Cavagna et al., 1964; Mayhew, 1977; Carrier, 1984; Brueckner et al., 1991; Bourdin et al., 1993; di

Prampero et al., 1993; Dalleau et al., 1998; Sparrow, 2000; Wickler et al., 2000; McArdle et al., 2001; Bramble et al., 2004; Formenti et al., 2005; Abe et al., 2007; Snyder et al., 2008).

In other words, human running is not characterized by a speed at which running is optimally efficient (Capelli et al., 2008; Steudel-Numbers et al., 2009).

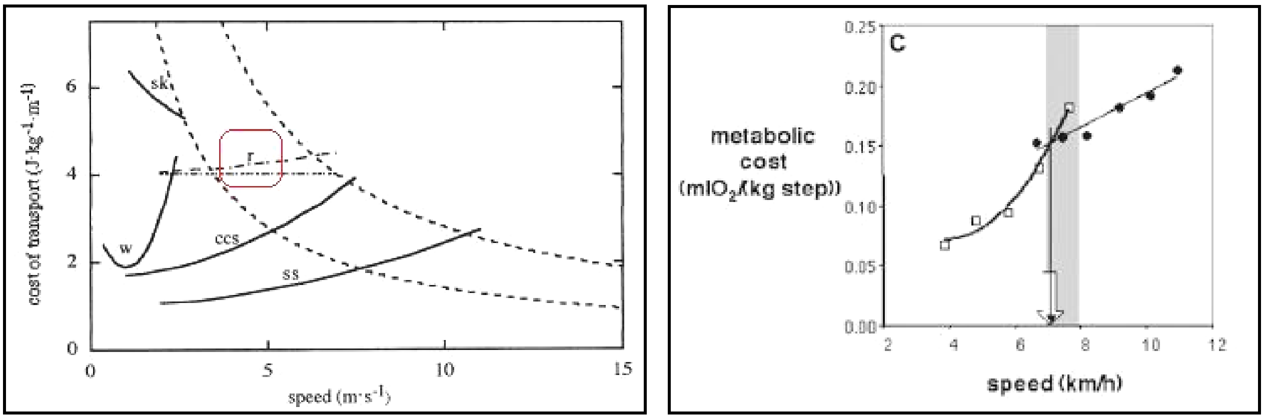


Figure 17.21. The relationship between the cost of progression and speed in different sports, in Saibene et al. (2003) (on the left). The relationship between metabolic cost and speed (walking and running), in Saibene et al. (2003) (on the right).

The Figure 17.22 shows how the independence between metabolic cost and running speed could be observed in different trained runners (average value = 3.5 or 4 J/(kg·m)).

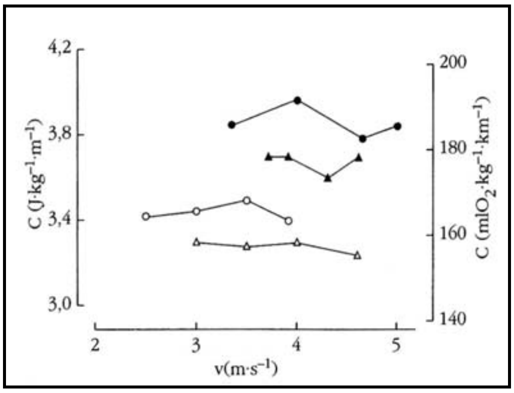


Figure 17.22. The linear relationship between the metabolic cost and speed in running, in di Prampero et al. (1993).

The independence between metabolic cost and running speed has been verified both in level and in grade (uphill or downhill) running (Figure 17.23; di Prampero, 1985; di Prampero et al., 2002; Saibene et al., 2003; Capelli et al., 2008; di Prampero et al., 2009).

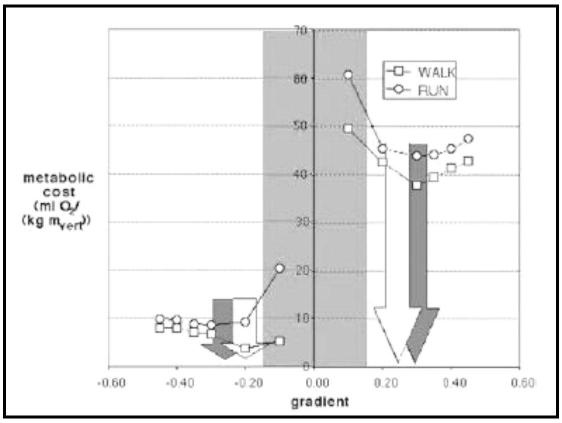


Figure 17.23. The relationship between vertical metabolic cost and gradient (walking and running), in Saibene et al. (2003).

However, in their pilot study, Margaria et al. (1963) took indirect calorimetric measurements on two athletes running at different speeds up to 6.11 m/s at gradients from -20 to 15%: the function was found to be linearly related to speed. Within these limits, the net metabolic cost values seem to be independently of speed and related only to the gradient.

3.4. Metabolic cost in different trained runners

Over years, in literature, there have been many researches investigating the pattern of metabolic cost/running economy in runners featuring different abilities. It would be very long and difficult to try to recognize and summarize all these studies. Consequently, only the most important research projects have been presented.

However, for more information and details, see also chapter 18 (par. 3.3).

In the past, elite and good distance runners were found to be significantly different in oxygen uptake during a standardized sub-maximal treadmill running (Cavanagh et al., 1977). Furthermore, it appeared to be a substantial variation in oxygen consumption at the sub-maximal running speed between trained and untrained runners (Mayhew, 1977). Moreover, improvements in running efficiency resulting from training have been shown to be 0-8%, although some research had indicated that it can be as high as 50-54% (Dill, 1965). Importantly, distance athletes ran 5 to 10% more economically than well-trained middle-distance runners (Conley et al., 1980). However, no significant differences in the net energy cost during running between similarly trained groups of men and women have been found (Bunc et al., 1989; Daniels et al., 1992). This similarity has been explained by the similar training states of both sexes, resulting from the intense training which did not differ in its relative intensity and frequency between the groups of men and women.

Quite recently, it has been demonstrated as multiple trials are not required to obtain stable measures of running economy and basic mechanical characteristics in trained male runners if group data from adequate sample sizes are considered. However, if individual records are scrutinized or if small sample sizes cannot be avoided, at least two measures of individual performances should be secured (Morgan et al., 1991). Furthermore, the influence of body dimensions on running economy comparing athletes specialized in different competition events has been studied (Maldonado et al., 2002). These data suggested that highly trained distance runners tend to show counterbalancing profiles of running economy and $V'O_{2max}$ (the higher the energy cost, the higher the $V'O_{2max}$ and *vice versa*), and that anthropometric characteristics related with good performance are different in long-distance and middle-distance events. In another study, the main differences in $V'O_{2max}$ and $V'O_{2submax}$ in various trained runners have been investigated (Morgan et al., 1995). In detail, R1 represents more trained subjects; R2 and R3 middle trained runners; and R4 less trained runners. Their results show that: a) average $V'O_{2max}$ slightly decreases as a function of training, while b) $V'O_{2submax}$ increases (R1 > R2 > R3 > R4). Consequently, the least training group is more expensive compared to the other runners, whereas the most trained runners are the most economical. To sum up, metabolic cost linearly decreases with training.

3.5. How we calculated metabolic cost of running

Instead of many different methods to estimate the metabolic cost proposed in both human walking and running (Andolf et al., 1976; Zarrugh et al., 1978; Waters et al., 1983; Brockway, 1987; Sherman, 1998; Waters et al., 1999; Detrembleur et al., 2003; Biewener et al., 2004; Abe et al., 2007; van de Hecke et al., 2007; Schwartz, 2007; van Engelen et al., 2009; Gordon et al., 2009; Mahaudens et al., 2009; Massaad et al., 2009; di Prampero et al., 2009; Plasschaert et al., 2009a; 2009b), according to literature (see par. 3.2 above), the Equation we have then used to express C was [17.7b]. In fact, in our analysis, the metabolic rate was determined from oxygen consumption ($V'O_2$) measured by indirect calorimetry (see also chapter 15, par. 4.1). The analyser, calibrated prior to each testing session, provided breath-by-breath data sent via telemetry to a computer (see also chapter 15, par. 4.3).

Firstly, the average of the final 2 minutes of sampling during the rest period was used for further analysis (Jones et al., 1996; Lejeune et al., 1998; Millet et al., 2000; Minetti et al., 2001; Millet et al., 2003; Doke et al., 2004; Ardigò et al., 2005; Gottschall et al., 2005; Modica et al., 2005; Mian et al., 2006; Minetti et al., 2006; Abe et al., 2007; Ortega et al., 2007; Teunissen et al., 2007; Bar-Haim et al., 2008; Capelli et al., 2008; Zamparo et al., 2008; Peyrot et al., 2009).

Secondly, the average of the final 2 minutes of sampling at each running speed was used for further analysis (Donelan et al., 2002). At each speed, this averaging phase did not start until the conditions described in chapter 16 (par. 2) had been met.

Importantly, net V’O_2 (obtained by subtracting pre-exercise standing V’O_2 from gross V’O_2) was converted to joules using an energetic equivalent of ≈ 20.9 (J/(kg·m)) O_2 (i.e. (4.94·RER)+16.04)): see Equation [17.7b] and [17.8b]. This procedure concurs with di Prampero (1985; 1986); Blaxter (1989); Lejeune et al. (1998); Minetti et al. (2001); Alexander (2003); Doke et al. (2004); Ardigò et al. (2005); Beneke et al. (2005); Gottschall et al. (2005); Modica et al. (2005); Mian et al. (2006); Minetti et al. (2006); Ortega et al. (2007); Schwartz (2007); Capelli et al. (2008); Zamparo et al. (2008); Mahaudens et al. (2008; 2009); McGregor et al. (2009); Nakai et al. (2009); Peyrot et al. (2009); Plasschaert et al. (2009a).

Finally, C was then obtained by dividing net energy expenditure (J/(kg·m)) by speed (m/s), according to Equation [17.7b].

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PART 2

RESULTS AND DISCUSSION

Chapter 18

BODILY SYMMETRIES AND RUNNING ECONOMY RESULTS

1. STATIC ANATOMICAL SYMMETRIES: TWO-DIMENSIONAL ANALYSIS

1.1. Introduction

In the following sections, the main results obtained by applying a two-dimensional analysis have been reported and discussed. In order to make easy their understanding, the distinction in the three segments (foot segment and shank area, knee joint and thigh area and pelvic region) has been maintained (see chapter 17, par. 1.2.2).

One MRI in an occasional runner and MRI in three top runners have to be excluded of our analysis because of a low imaging resolution. So that, 15 MRI were considered at all.

1.2. Results of our study

1.2.1. Foot segment and shank area

Average values of Static Symmetry Ratio (and their standard deviation) in each foot segment and shank area are presented in Table 18.1 (occasional, skilled and top runners).

ANATOMICAL REGION	OCCASIONAL RUNNERS		SKILLED RUNNERS		TOP RUNNERS	
	Mean ± S.D.	**CV (%)**	**Mean ± S.D.**	**CV (%)**	**Mean ± S.D.**	**CV (%)**
L1	0.94 ± 0.07	7.4	1.05 ± 0.10	9	1.03 ± 0.10	10.2
L2	1.03 ± 0.09	9	1.03 ± 0.08	8	1.02 ± 0.13	12.8
L3	1.01 ± 0.08	7.7	1.03 ± 0.12	12	1.18 ± 0.16	13.9
L4	0.97 ± 0.11	11	1.02 ± 0.04	4	0.93 ± 0.07	7.1
L5	1.01 ± 0.09	8.4	1.01 ± 0.04	4	1.05 ± 0.04	3.6
L6	1.04 ± 0.11	11	1.03 ± 0.11	11	1.10 ± 0.06	5.7
L7	0.97 ± 0.08	9	0.99 ± 0.03	3	0.96 ± 0.00	0.4
%L8	0.96 ± 0.05	5	1.02 ± 0.10	10	1.05 ± 0.02	2.1
%L9	1.03 ± 0.10	10	0.99 ± 0.04	4	0.91 ± 0.04	4
%L10	1.03 ± 0.10	10	0.97 ± 0.15	15	1.04 ± 0.02	2.2

Table 18.1. Average value (± S.D.) and CV (%) of the left/right mathematical ratio, in the leg segment.

Moreover, single Static Symmetry Ratio values for each area (and the corresponding difference between the optimal SSR - 1 - and each single value) are illustrated in Table 18.2a, 18.2b and 18.2c in occasional, skilled and top runners, respectively.

FOOT SEGMENT AND SHANK AREA	SSR	|1-SSR|	MAJOR AREA
L1	0.94	0.06	Right
L2	1.03	0.03	Left
L3	1.01	0.01	Left
L4	0.97	0.03	Right
L5	1.01	0.01	Left
L6	1.04	0.04	Left
L7	0.97	0.03	Right
%L8	0.96	0.04	Right
%L9	1.03	0.03	Left
%L10	1.03	0.03	Left

Table 18.2a. All SSR values of foot segment and shank area, occasional runners.

FOOT SEGMENT AND SHANK AREA	SSR	|1-SSR|	MAJOR AREA
L1	1.05	0.05	Left
L2	1.03	0.03	Left
L3	1.03	0.03	Left
L4	1.02	0.02	Left
L5	1.01	0.01	Left
L6	1.03	0.03	Left
L7	0.99	0.01	Right
%L8	1.02	0.02	Left
%L9	0.99	0.01	Right
%L10	0.97	0.03	Right

Table 18.2b. All SSR values of foot segment and shank area, skilled runners.

FOOT SEGMENT AND SHANK AREA	SSR	|1-SSR|	MAJOR AREA
L1	1.03	0.03	Left
L2	1.02	0.02	Left
L3	1.18	0.18	Left
L4	0.93	0.07	Right
L5	1.05	0.05	Left
L6	1.10	0.10	Left
L7	0.96	0.04	Right
%L8	1.05	0.05	Left
%L9	0.91	0.09	Right
%L10	1.04	0.04	Left

Table 18.2c. All SSR values of foot segment and shank area, top runners.

A one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) were applied to investigate whether significant differences there will between occasional and skilled groups. Yet, because of the poor number of investigated subjects (n = 2; indeed, three MRI were rejected), top runners cannot be involved in this statistical analysis. Therefore, in this last case, only a qualitative approach has been applied.

The following figures contain and show the graphs corresponding to each foot and shank area, in the single runners group. Precisely, in each graph, the histograms represent mean values obtained by grouping the same running levels (OR *versus* SR *versus* TR) and the vertical bars the positive standard deviations (mean ± S.D.). Particularly:

- red corresponds to occasional runners’ specific anatomical region;
- green corresponds to skilled runners’ specific anatomical region;
- ski-blue corresponds to top runners’ specific anatomical region.

Each graph ranges from 0.0 to 1.4 (step 0.2), according to what explained in chapter 17, par. 1.2.6. Significances have been highlighted by the asterisks.

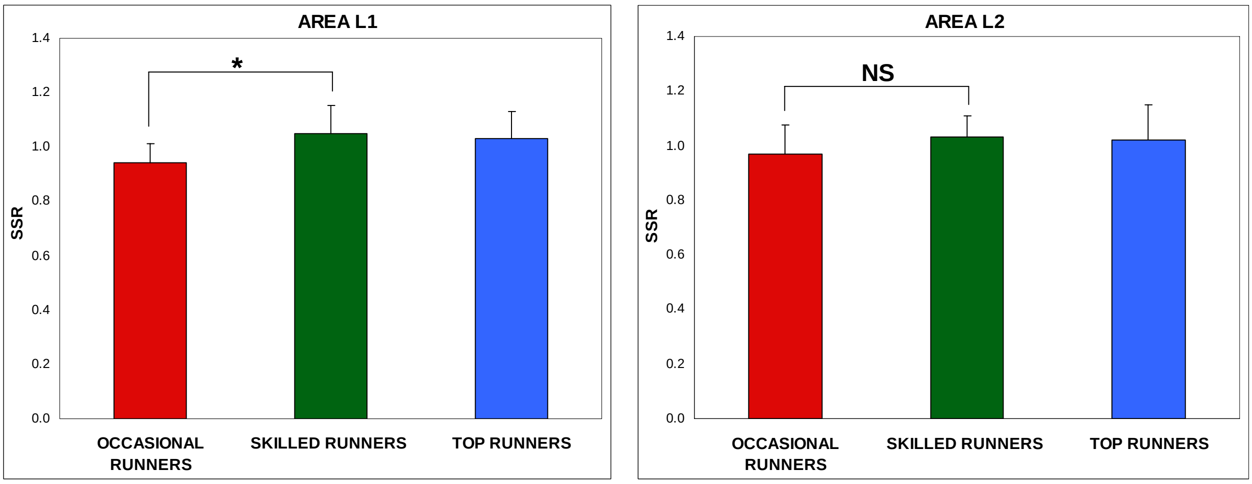


Figure 18.1. Static Symmetry Ratio in heel area (on the left) and in heel bone area (on the right).

1. In <u>heel area</u> (L1), occasional runners seem to have a higher right symmetry compared to both skilled ($p<0.05$) and top runners. Moreover, the qualitative analysis has shown that skilled and top runners are quite similar.

2. In <u>heel bone area</u> (L2), both the statistical and the qualitative analysis have shown that there are no significant differences among runners.

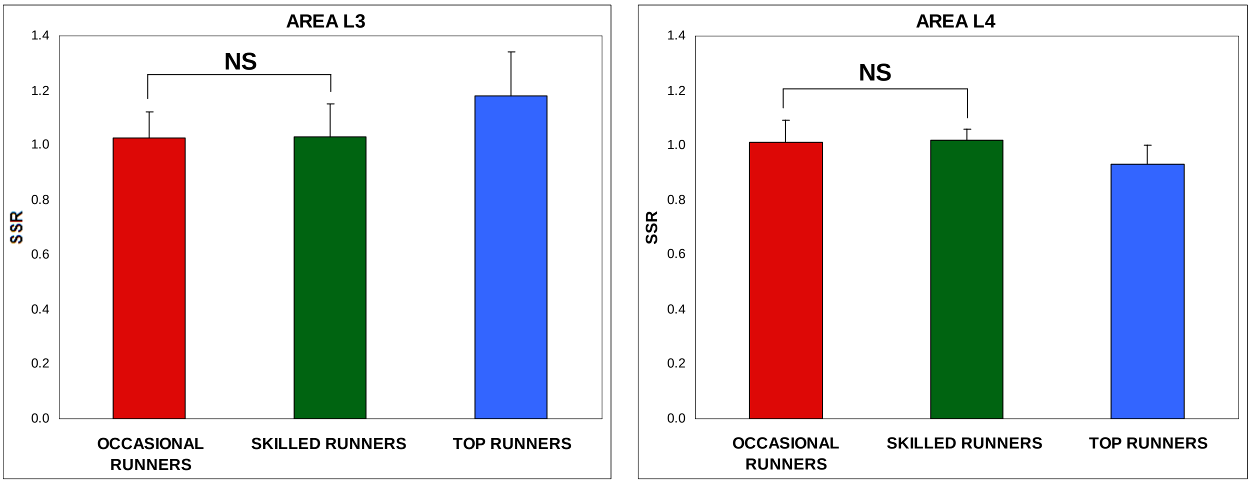


Figure 18.2. Static Symmetry Ratio in heel area when fibulae are (on the left)
and are not (on the right) shown.

3. In heel area when fibulae are shown (L3), the statistical analysis has shown that there are no significant differences between occasional and skilled runners. Differently, top runners seem to have a higher left area compared to the other groups. In this last case, the standard deviation is higher, as well.

4. In heel area when fibulae are not shown (L4), the statistical analysis has shown that there are no significant differences between occasional and skilled runners. Differently, top runners seem to have a slightly higher right area.

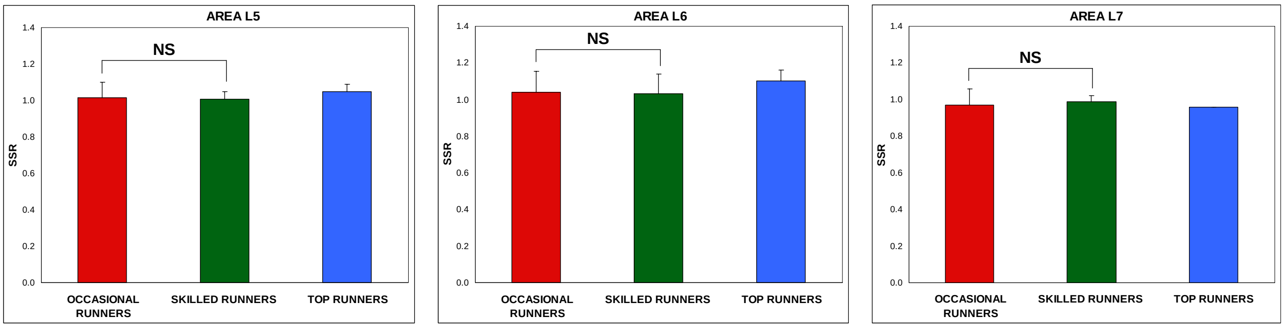


Figure 18.3. Static Symmetry Ratio in total shank area (on the left), in shank muscle area (in the middle) and in shank bone area (on the right).

5. In total shank area (L5), both the statistical and the qualitative analysis have shown that there are no significant differences among runners.

6. In shank muscles area (L6), the statistical analysis has shown that there are no significant differences between occasional and skilled runners. Differently, top runners seem to have a higher left area.

7. In shank bones area (L7), both the statistical and the qualitative analysis have shown that there are no significant differences among runners.

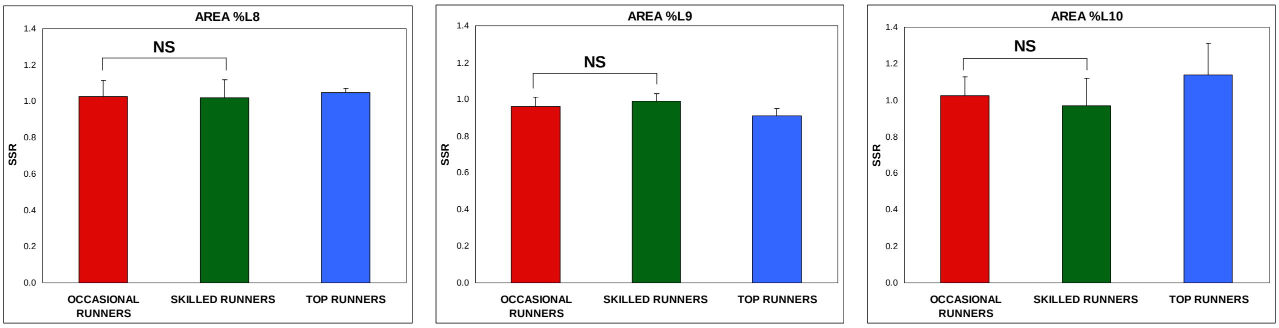


Figure 18.4. Static Symmetry Ratio in shank muscle percentage (on the left), in shank bone percentage (in the middle) and in shank other tissues percentage (on the right).

8. In shank muscles percentage (%L8), both the statistical and the qualitative analysis have shown that there are no significant differences among runners.

9. In shank bones percentage (%L9), the statistical analysis has shown that there are no significant differences between occasional and skilled runners. Differently, top runners seem to have a slightly higher right area.

10. In shank other tissues percentage (%L10), the statistical analysis has shown that there are no significant differences between occasional and skilled runners. Differently, top runners seem to have a slightly higher left area. In all these cases, the standard deviation is higher, as well.

In conclusion, our results show that there are very little differences among runners when the foot and shank segments are investigated, in both their area and percentage.

Single runner MRI (and relative Static Symmetry Ratio) are contained in the enclosed CD (Second Study, Chapter 18, MRI all runners).

1.2.2. Knee joint and thigh area

Average values of Static Symmetry Ratio (and their standard deviation) in each knee joint and thigh area are presented in Table 18.3 (occasional, skilled and top runners).

ANATOMICAL REGION	OCCASIONAL RUNNERS		SKILLED RUNNERS		TOP RUNNERS	
	Mean ± S.D.	**CV (%)**	**Mean ± S.D.**	**CV (%)**	**Mean ± S.D.**	**CV (%)**
T1	0.90 ± 0.10	12.4	1.04 ± 0.05	4	1.25 ± 0.02	0.01
T2	0.98 ± 0.06	6	0.98 ± 0.06	6	1.04 ± 0.03	0.03
T3	0.99 ± 0.03	3	0.99 ± 0.04	4	1.05 ± 0.01	0.01
T4	1.03 ± 0.02	2	0.98 ± 0.05	5	1.08 ± 0.02	0.02
%T5	1.03 ± 0.03	3	0.99 ± 0.01	2	1.03 ± 0.01	0.01
%T6	0.91 ± 0.11	11.6	1.05 ± 0.08	7	1.18 ± 0.02	0.02
%T7	1.03 ± 0.06	6	1.00 ± 0.06	6	0.78 ± 0.00	0.00

Table 18.3. Average value (± S.D.) and CV (%) of the left/right mathematical ratio, in the thigh segment.

Moreover, single Static Symmetry Ratio values for each area (and the corresponding difference between the optimal SSR and each single value) are illustrated in Table 18.4a, 18.4b and 18.4c in occasional, skilled and top runners, respectively.

KNEE JOINT AND THIGH AREA	SSR	|1-SSR|	MAJOR AREA
T1	0.90	0.10	Right
T2	0.98	0.02	Right
T3	0.99	0.01	Right
T4	1.03	0.03	Left
%T5	1.03	0.03	Left
%T6	0.91	0.09	Right
%T7	1.03	0.03	Left

Table 18.4a. All SSR values of knee joint and thigh area, occasional runners.

KNEE JOINT AND THIGH AREA	SSR	|1-SSR|	MAJOR AREA
T1	1.04	0.04	Left
T2	0.98	0.02	Right
T3	0.99	0.01	Right
T4	0.98	0.02	Right
%T5	0.99	0.01	Right
%T6	1.05	0.05	Left
%T7	1.00	0.00	Right = Left

Table 18.4b. All SSR values of knee joint and thigh area, skilled runners.

KNEE JOINT AND THIGH AREA	SSR	|1-SSR|	MAJOR AREA
T1	1.25	0.25	Left
T2	1.04	0.04	Left
T3	1.05	0.05	Left
T4	1.08	0.08	Left
%T5	1.03	0.03	Left
%T6	1.18	0.18	Left
%T7	0.78	0.22	Right

Table 18.4c. All SSR values of knee joint and thigh area, top runners.

The following figures contain and show the graphs corresponding to each knee joint and thigh area, in the single runners group. The same legend described in par. 1.2.1 has been used.

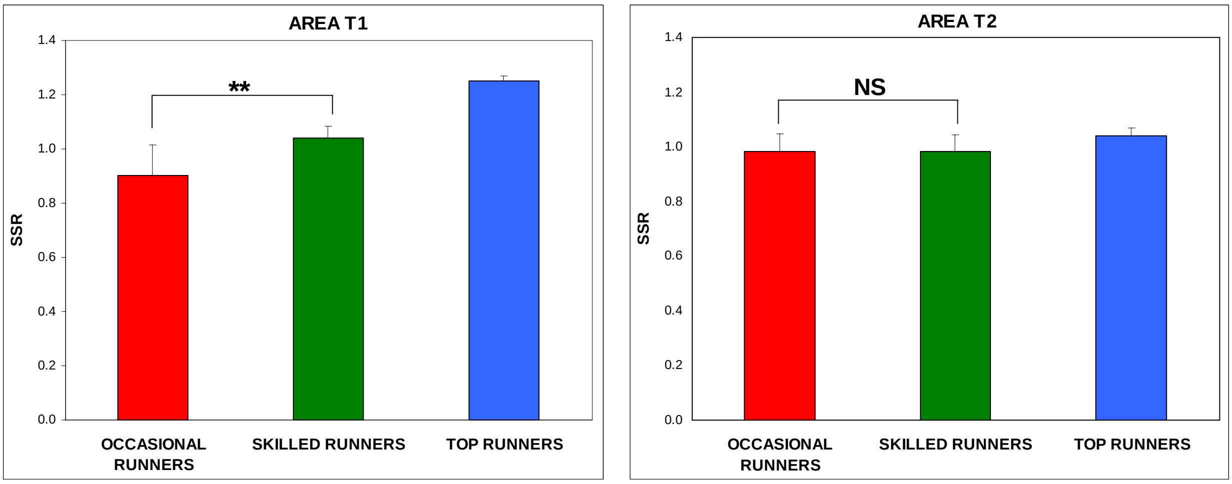


Figure 18.5. Static Symmetry Ratio in long bone area when femur (on the left) and femur and tibiae (on the right) are shown.

1. In long bone area when femur has shown (T1), the statistical analysis has shown that skilled runners have a higher left symmetry compared to occasional runners ($p<0.01$). Furthermore, it seems that top runners have the most left area.

2. In long bone area when femur and tibiae have shown (T2), both the statistical and the qualitative analysis have shown that there are no differences among runners.

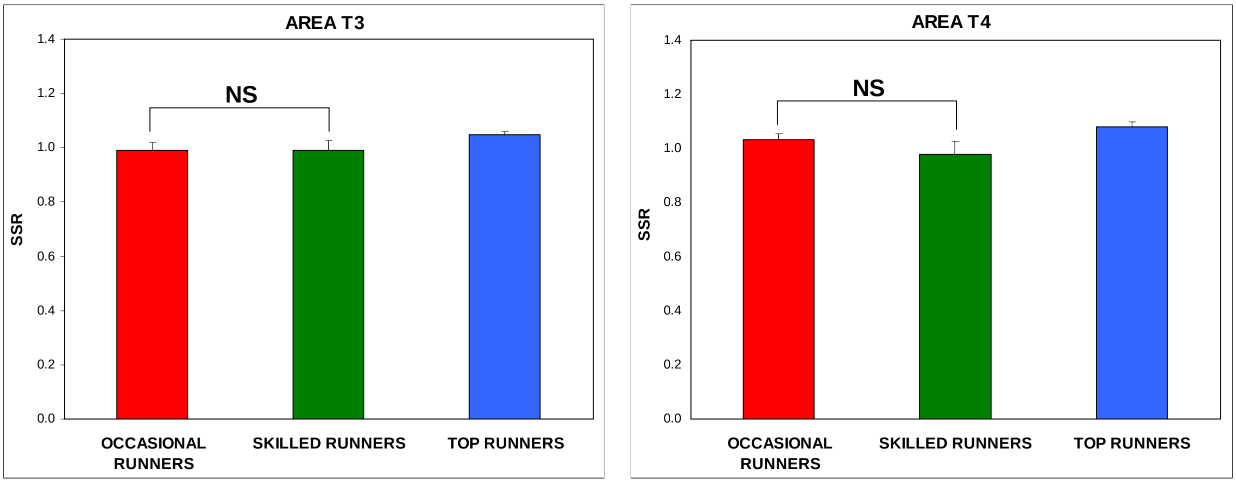


Figure 18.6. Static Symmetry Ratio in total thigh area (on the left) and in thigh muscle area (on the right).

3. In total thigh area (T3), the statistical analysis has shown that there are no differences between occasional and skilled runners. Differently, it seems that top runners have a slightly higher left area.

4. In thigh muscles area (T4), the statistical analysis has shown that there are no differences between occasional and skilled runners. Differently, it seems that top runners have a slightly higher left area.

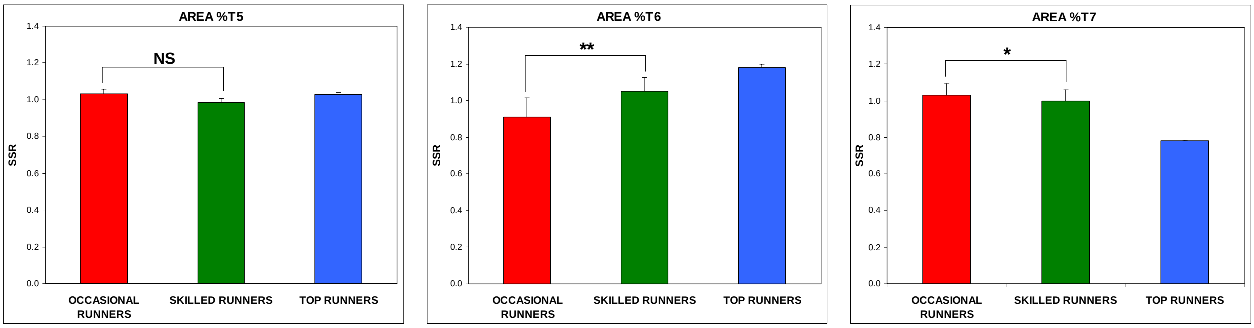


Figure 18.7. Static Symmetry Ratio in thigh muscle percentage (on the left), in thigh bone percentage (in the middle) and in thigh other tissues percentage (on the right).

5. In thigh muscles percentage (%T5), both the statistical and the qualitative analysis have shown that there are no differences among runners.

6. In thigh bones percentage (%T6), both the statistical and the qualitative analysis have shown that there are significant differences among runners. Indeed, occasional runners have a higher right area compared to skilled runners ($p<0.01$), whereas skilled runners have a higher right area compared to top runners.

7. In thigh other tissues percentage (%T7), the statistical analysis has shown that there are significant differences between occasional and skilled runners. Indeed, occasional runners have a higher left area compared to skilled runners (p<0.05). Differently, top runners seem to have the highest right area compared to the other groups.

In conclusion, our results show that there are only little differences among runners when the knee joint and thigh segments are investigated, in both their area and percentage.

Single runner MRI (and relative Static Symmetry Ratio) are contained in the enclosed CD (Second Study, Chapter 18, MRI all runners).

1.2.3. Hip joint and pelvic area

Because of the already discussed absence of such analysis in this anatomical region (see chapter 17, par. 1.2.5), results are not available.

1.3. Discussion

As regards both foot and shank area and knee joint and thigh area, occasional runners have been compared to skilled runners. In this way, it has been possible to highlight the main differences (and their significance) for each group, in each single area, as shown in Table 18.5a (foot and shank area) and 18.5b (knee joint and thigh area). However, because the poor number of tested subjects, top runners were not involved in this analysis.

A one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) were used to state the differences between each runner group. The alpha test level set for statistical significance was 0.05. In the next tables:

a) NS is used if there is no significance;
b) * if there is a little significance (p<0.05);
c) ** if there is a good significance (p<0.01);
d) *** in the case of the highest significance (p<0.001).

AREA of INTEREST	**OR *versus* SR**	**OR *versus* TR**	**SR *versus* TR**
L1	*	result no available	result no available
L2	p=NS		
L3	p=NS		
L4	p=NS		
L5	p=NS		
L6	p=NS		
L7	p=NS		
%L8	p=NS		
%L9	p=NS		
%L10	p=NS		

Table 18.5a. Comparison among runner groups in foot segment and shank area.

AREA of INTEREST	**OR *versus* SR**	**OR *versus* TR**	**SR *versus* TR**
T1	**	result no available	result no available
T2	p=NS		
T3	p=NS		
T4	p=NS		
%T5	p=NS		
%T6	**		
%T7	*		

Table 18.5b. Comparison among runner groups in knee joint and thigh area.

In foot and shank area, there are no significant differences between occasional and skilled runners excepted in area L1, whereas, in knee joint and thigh area, the differences among groups are in T1, %T6 and %T7.

Furthermore, each runner group (inclusive top runners) has been also compared to what is optimal symmetry ratio in which SSR equals 1 (i.e. there are no differences between the left and the right side). A one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) were used to state the differences between each runner group and the optimal condition (SSR = 1). The alpha test level set for statistical significance was 0.05.

Single values used in statistical analysis are contained in the enclosed CD (Second Study, Chapter 18, MRI all runners: statistical analysis).

Table 18.6a shows foot segment and shank area results, whereas Table 18.6b shows knee joint and thigh area results, as well.

AREA of INTEREST	**OR *versus* 1**	**SR *versus* 1**	**TR *versus* 1**
L1	* (Right)	p=NS (Left)	p=NS (Left)
L2	p=NS (Left)	p=NS (Left)	p=NS (Left)
L3	p=NS (Left)	p=NS (Left)	*** (Left)
L4	p=NS (Right)	p=NS (Left)	* (Right)
L5	p=NS (Left)	p=NS (Left)	p=NS (Left)
L6	p=NS (Left)	p=NS (Left)	** (Left)
L7	p=NS (Right)	p=NS (Right)	p=NS (Right)
%L8	p=NS (Right)	p=NS (Left)	p=NS (Left)
%L9	p=NS (Left)	p=NS (Right)	** (Right)
%L10	p=NS (Left)	p=NS (Right)	p=NS (Left)

Table 18.6a. Comparison between the SSR in each group and the optimal symmetry ratio, in foot segment and shank area.

In occasional and skilled runners, there are very little significance if the experimental value of Static Symmetry Ratio has been compared to the optimal condition ($p<0.05$ in L1). This result suggests that, in these groups, right and left static symmetry are so close and similar, in each area and percentage. However, in top runners, a slight significance could be observed in some areas and percentages ($p<0.001$ in L3; $p<0.05$ in L4; $p<0.01$ in L6; and $p<0.01$ in %L9). This important result is probably due to an increased static asymmetry between both the right and the left side.

AREA of INTEREST	OR *versus* 1	SR *versus* 1	TR *versus* 1
T1	** (Right)	* (Left)	*** (Left)
T2	p=NS (Right)	p=NS (Right)	p=NS (Left)
T3	p=NS (Right)	p=NS (Right)	p=NS (Left)
T4	p=NS (Left)	p=NS (Right)	* (Left)
%T5	p=NS (Left)	p=NS (Right)	p=NS (Left)
%T6	** (Right)	* (Left)	*** (Left)
%T7	* (Left)	* (Left)	*** (Right)

Table 18.6b. Comparison between the SSR in each group and the optimal symmetry ratio, in knee joint and thigh area.

In occasional runners there are some significances if the experimental value of Static Symmetry Ratio has been compared to the optimal condition ($p<0.01$ in T1 and %T6; and $p<0.05$ in %T7). In all these regions, this result suggests that these subjects seem to have an increased right symmetry. Moreover, in skilled runners, there are some little significance ($p<0.05$ in T1; %T6 and %T7). Differently to what previously demonstrated in occasional runners, skilled runners also seem to have an increased left symmetry. Finally, top runners show significant differences if their average value has been compared to the optimal condition ($p<0.001$ in T1; $p<0.05$ in T4; and $p<0.001$ in %T6 and %T7). This important result shows that top runners are the most asymmetrical subjects in static condition. In particular, they have a higher left area, as well.

Overall, because of this poor and no-homogenous significance, we could state the **lack of consistency** in these results. Indeed, this bi-dimensional approach seems to be not so precise, valid and available to find out symmetry differences (or similarities) in runner groups.

2. STATIC ANATOMICAL SYMMETRIES: THREE-DIMENSIONAL ANALYSIS

2.1. Introduction

In the following sections, the main results obtained by applying a three-dimensional mathematical analysis have been reported and discussed.

Data refers to average cross-correlation coefficients or static anatomical indexes (see chapter 17, par. 1.3.2). On whole, only 15 MRI were considered (see par. 1.1 above). On average, these indexes range from 0.7 to 1 (step 0.05). Clearly:

- 0.7 corresponds to the lowest correspondence between the right and the left side;
- 1 corresponds to the optimal match.

2.2. Results of our study

Three-dimensional analysis mainly investigated three areas: ankle, knee and femur. Therefore, the static anatomical cross-correlation coefficients (and their standard deviation) related to these areas have been presented in Table 18.7a:

ANATOMICAL REGION	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
ANKLE AREA	0.890 ± 0.036	0.892 ± 0.039	0.894 ± 0.037
KNEE AREA	0.893 ± 0.040	0.894 ± 0.053	0.888 ± 0.014
FEMUR AREA	0.852 ± 0.057	0.808 ± 0.090	0.860 ± 0.045

Table 18.7a. Average static anatomical indexes in ankle, knee and femur areas.

A one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) were performed to compare the main effects of both anatomical region and group. The alpha test level set for statistical significance was 0.05.

Our results show that:

- in ankle and knee areas, there are no significant differences as a function of running ability. However, in femur area, average cross-correlation indexes are significantly lower in skilled runners ($p<0.001$), while there are no differences among other groups (Table 18.7b);

RUNNERS	ANKLE AREA	KNEE AREA	FEMUR AREA
OCCASIONAL RUNNERS	p=NS	p=NS	$p<0.001$ between OR and SR $p<0.001$ between SR and TR p=NS between OR and TR
SKILLED RUNNERS			
TOP RUNNERS			

Table 18.7b. Comparison of cross-correlation coefficients in ankle, knee and femur areas as a function of running ability.

- on average, in the femur area, the correspondence between the right and the left side is the lowest compared to other areas: $p<0.01$ in both occasional and top runners and $p<0.001$ in skilled runners (Table 18.7c).

ANATOMICAL REGION	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
ANKLE and KNEE AREA	$p<0.01$ by comparing femur area to other areas	$p<0.001$ by comparing femur area to other areas	$p<0.01$ by comparing femur area to other areas
FEMUR AREA			

Table 18.7c. Comparison of cross-correlation coefficients in single runner groups as a function of ankle, knee and femur areas.

In addition, static anatomical cross-correlation coefficients have been measured in lower and upper leg, as well. These values (and their standard deviation) have been presented in Table 18.8a:

ANATOMICAL REGION	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
LOWER LEG	0.832 ± 0.058	0.847 ± 0.056	0.809 ± 0.015
UPPER LEG	0.840 ± 0.035	0.805 ± 0.068	0.798 ± 0.021

Table 18.8a. Average static anatomical indexes in lower and upper leg.

A one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction) were performed to compare the main effects of both anatomical region and group. The alpha test level set for statistical significance was 0.05.

Our results show that:

- in lower leg area, there are significant differences as a function of running ability. Indeed, the best average cross-correlation index is in skilled runners, whereas the lowest one is in top runners. Furthermore, in upper leg area, the best average cross-correlation index is in occasional runners ($p<0.001$), while there are no differences among other groups (Table 18.8b);

RUNNERS	**LOWER LEG**	**UPPER LEG**
OCCASIONAL RUNNERS	$p<0.01$ between OR and SR $p<0.001$ between SR and TR $p<0.001$ between OR and TR	$p<0.001$
SKILLED RUNNERS		$p=\mathrm{NS}$
TOP RUNNERS		

Table 18.8b. Comparison of cross-correlation coefficients in lower and upper leg areas as a function of running ability.

- in occasional runners, average cross-correlation coefficients are slightly greater ($p<0.05$) in the upper leg. This means that the correspondence between the right and the left side is higher in this area. However, both in skilled and top runners, average cross-correlation coefficients are greater ($p<0.001$ and $p<0.01$, respectively) in the lower leg meaning that the correspondence between the right and the left side is higher in this area (Table 18.8c).

ANATOMICAL REGION	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
LOWER LEG **UPPER LEG**	$p<0.05$ (upper leg)	$p<0.001$ (lower leg)	$p<0.01$ (lower leg)

Table 18.8c. Comparison of cross-correlation coefficients in single runners as a function of lower and upper leg areas.

Single values in all runners are contained in the enclosed CD (Second Study, Chapter 18, MRI all runners: cross correlation coefficients).

2.3. Discussion

Independently of anatomical areas, the high cross-correlation coefficients mean that a great correspondence among sides is clearly marked. However, only slight differences are found as a function of both anatomical region and running ability.

Otherwise, the most important result is the high correspondence between the ankle and the knee area. In other words, a high static anatomical index in ankle area is coupled with a high index in knee area and *vice versa*, as shown in Figure 18.8 (n = 15, $R^2 = 0.5628$, r = 0.7502).

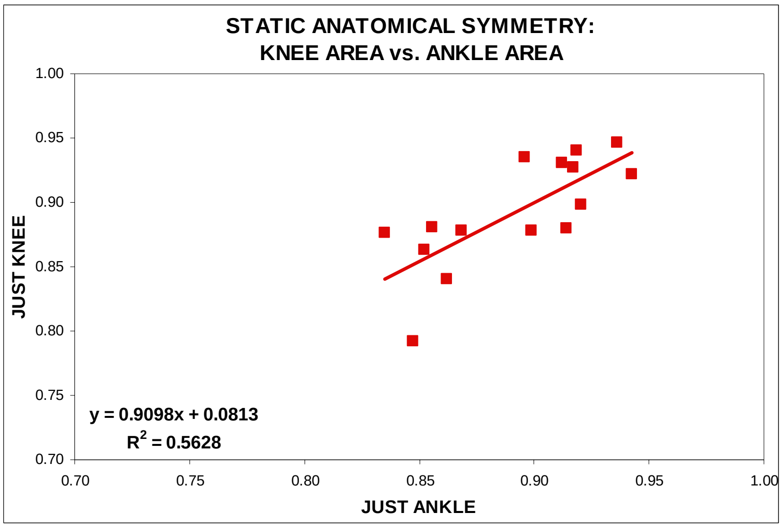


Figure 18.8. The high correspondence between ankle and knee areas.

Clearly, the close match between these areas highlights our hypothesis that a subject would be more symmetric independently of the investigated anatomical regions.

3. KINEMATIC FUNCTIONAL SYMMETRIES

3.1. Digital Locomotory Signature

Digital Locomotory Signature has been used to assess kinematic functional symmetries as a function of running speed. It has been calculated as widely described in chapter 6 and 7.

As already described in chapter 7 (par. 2), both a qualitative and a statistical analysis have been performed to compare average values of such continuous functions in runners of different ability.

Specifically, a one-way ANOVA with Huynh-Feldt correction for repeated measures were used to assess, in each movement direction, the main amplitude differences between occasional, skilled and top runners, and running speed (and the possible interaction among these two variables).

Finally, it is important to note that the previously described statistical analysis could not be applied to phase coefficients. Indeed, they constitute a circular variable differently to amplitudes (see also chapter 9, par. 2.2 and 2.3). In this case, a circular statistical analysis has to be performed. However, we have decided to consider the corresponding sine and cosine functions to solve this problem. Indeed, both sine and cosine are linear variables on which a one-way ANOVA with Huynh-Feldt correction for repeated measures could be applied in order to assess the main differences, between ability and speed, and the possible interaction among these two variables.

The last two amplitude coefficients (A5 and A6) were not considered in this statistical analysis because of their relative importance in the characterization of the Digital Locomotory Signature (see also chapter 7, par. 2.3). However, they were used in the graphical representation of the closed loops (qualitative analysis). For more details concerning this analysis, see chapter 7.

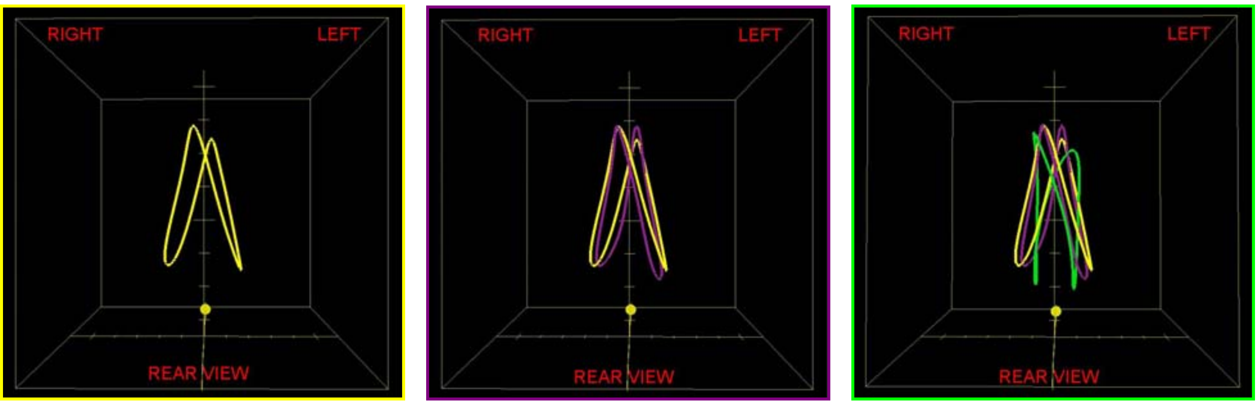
Single values of both symmetrical and asymmetrical coefficients (in runners) are contained in the enclosed CD (Second Study, Chapter 18, Coefficients statistical analysis).

In all the graphs below (which were drawn up using *Grapher*), yellow corresponds to the speed of 2.22 m/s; violet to 2.78 m/s; green to 3.33 m/s; red to 3.89 m/s; ski-blue to 4.44 m/s and white to 5.00 m/s.

Average 3D *contours* are contained in the enclosed CD (Second Study, Chapter 18, Grapher 3D contours).

3.1.1. Digital Locomotory Signature in occasional runners

In the graphs below, there are 3D *contours* in level running at each speed, in occasional runners (Figure 18.9: rear view; and Figure 18.10: front, lateral right and top views). Limits of each graph are: a) -30/30 mm along the forward direction; b) 890/1050 mm along the vertical direction; and c) -30/30 mm along the lateral direction.



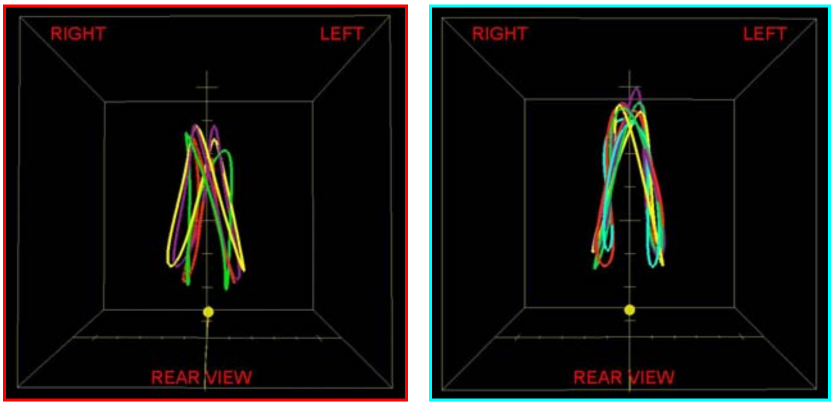


Figure 18.9. 3D *contours* in level running at all speeds (from 2.22 to 4.44 m/s), occasional runners (from the top left graph to the bottom right graph).

As shown in these *contours*, the BCOM raises and lifts as a function of running speed so that it becomes slightly more vertical (Lee et al., 2009). However, no more significant differences could be found.

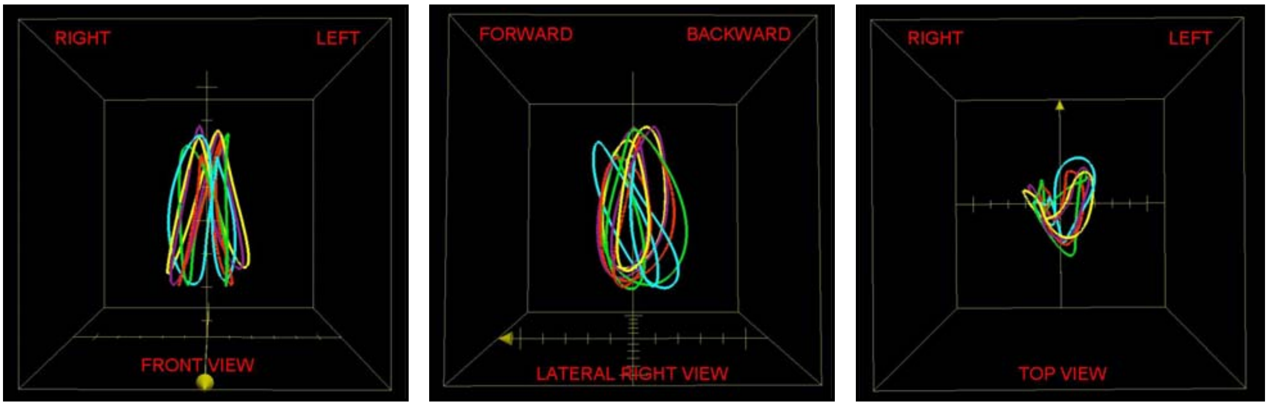


Figure 18.10. 3D *contours* in level running at all speeds, occasional runners (front view, lateral right view and top view).

The prospective views in Figure 18.10 above (the top view) emphasize the presence on a left asymmetry. It seems that this asymmetry slightly increases with speed (for instance, at 4.44 m/s the ski-blue loop is markedly bigger on the left side). This is probably related to the fact that 100% of occasional runners are right -footed and -handed (see also chapter 7, par. 3.2; Lund, 1930; Raibert, 1986; Song et al., 1997; Delattre et al., 2001; Souman et al., 2009; Strike et al., 2009).

3.1.2. Digital Locomotory Signature in skilled runners

In the graphs below, there are 3D *contours* in level running at each speed, in skilled runners (Figure 18.11 and 18.12). Limits of each graph are the same as occasional runners (see par. 3.1.1 above).

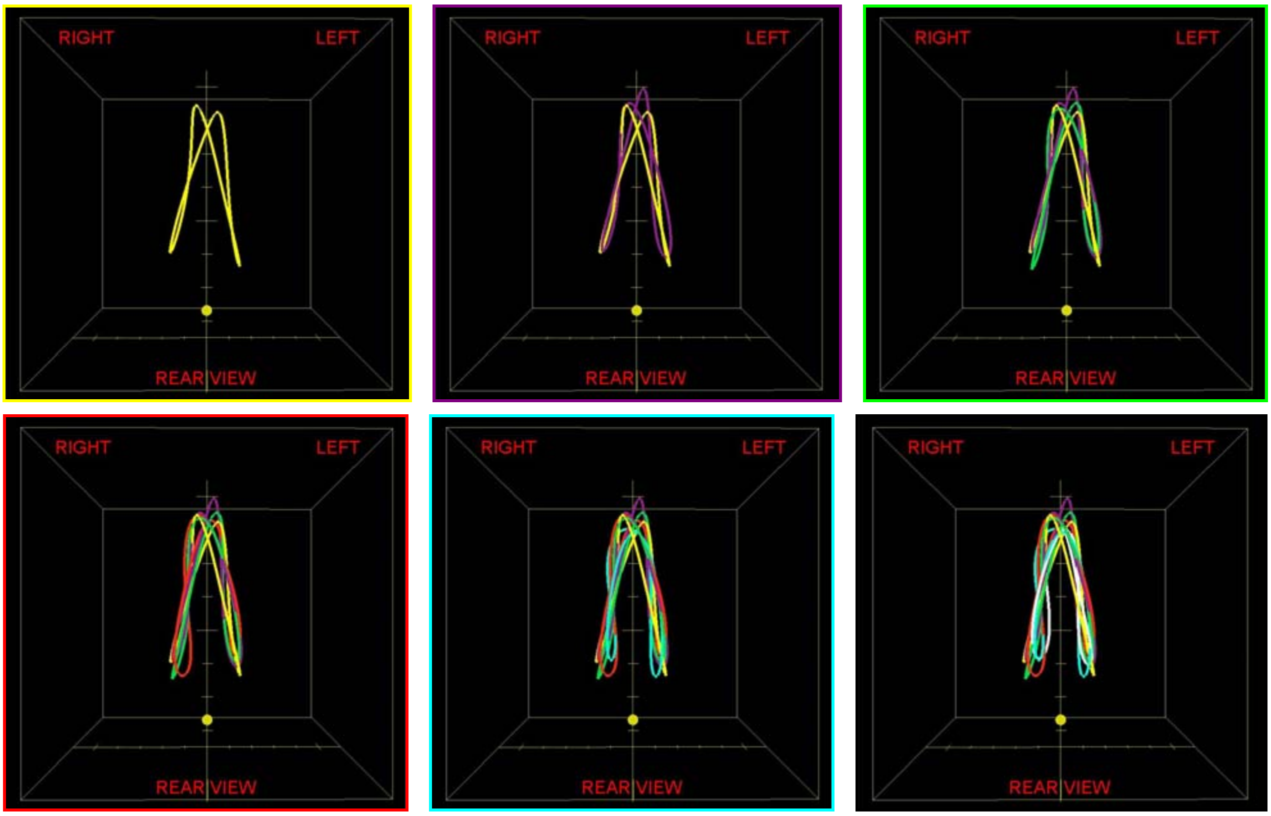


Figure 18.11. 3D *contours* in level running at all speeds (from 2.22 to 5.00 m/s), skilled runners (from the top left graph to the bottom right graph).

As already discussed in occasional runners, the BCOM raises and lifts as a function of running speed becoming slightly more vertical, as well (Lee et al., 2009).

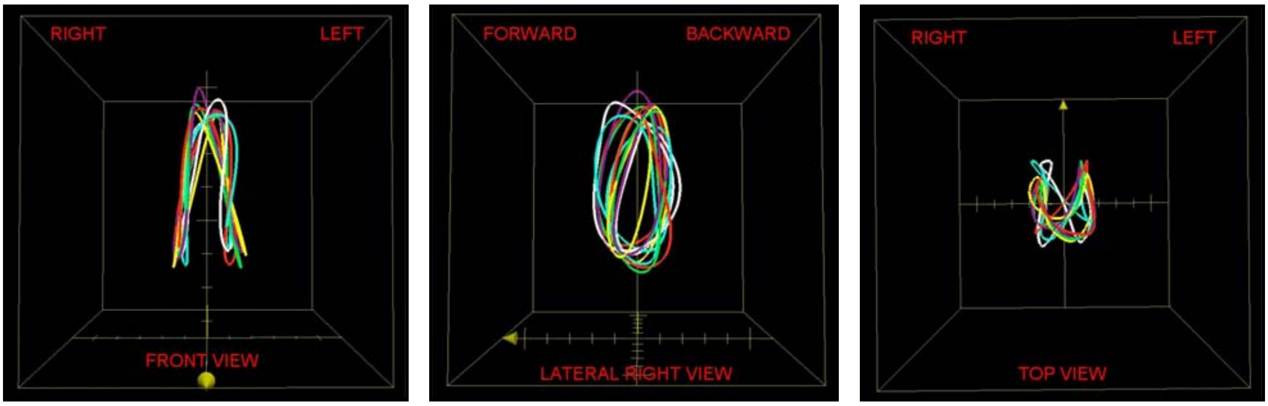


Figure 18.12. 3D *contours* in level running at all speeds, skilled runners (front view, lateral right view and top view).

The prospective views in Figure 18.12 above (the top view) emphasize that there are no so significant differences among speeds. Furthermore, the presence on a left asymmetry could be observed, too. Clearly, it is related to the fact that 100% of skilled runners are right -footed and -handed.

3.1.3. Digital Locomotory Signature in top runners

In the graphs below, there are 3D *contours* in level running at different speeds, in top runners (Figure 18.13 and 18.14). Limits of each graph are the same as occasional runners.

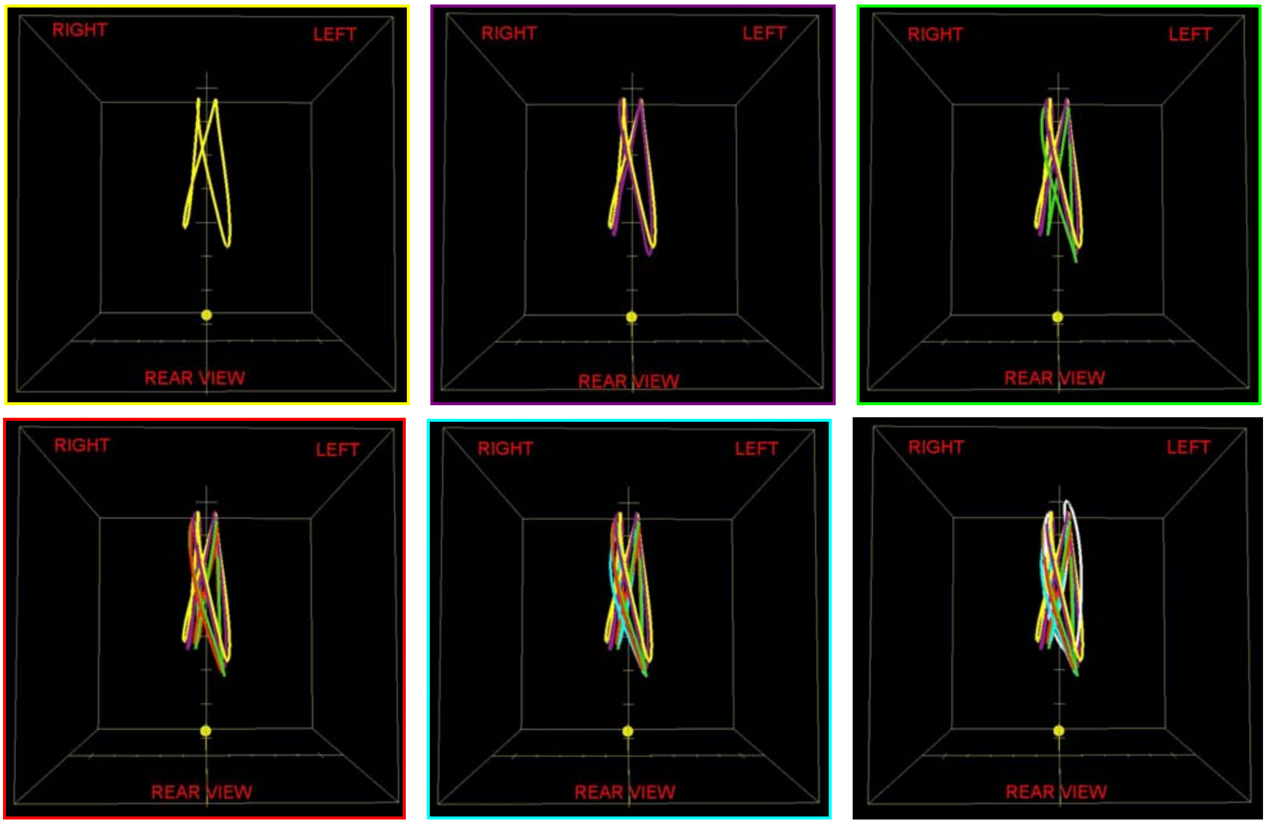


Figure 18.13. 3D *contours* in level running at all speeds (from 2.22 to 5.00 m/s), top runners (from the top left graph to the bottom right graph).

Contrarily to what previously discussed, in top runners, the raising and lifting of BCOM as a function of running speed is not so evident (i.e. the progressive vertical fall of anatomical markers).

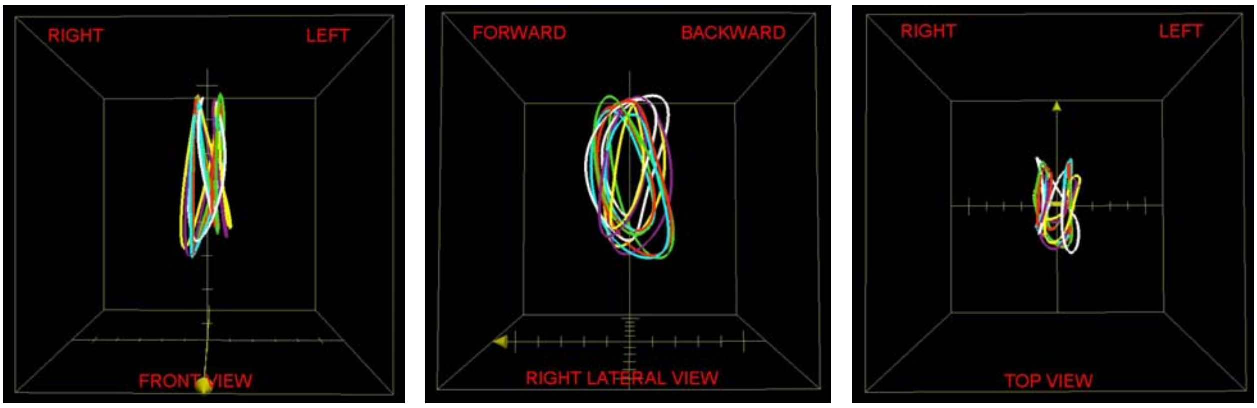


Figure 18.14. 3D *contours* in level running at all speeds, top runners (front view, lateral right view and top view).

However, the prospective views in Figure 18.14 above (the top view) reinforce the left asymmetry already described in all the other runners. It is related to the fact that 80% of top runners are right -footed and -handed.

3.1.4. Ability and speed: results of statistical analysis

A. In the forward direction, we come to the conclusion that: a) in all amplitudes and sine/cosine functions, all runners are similar; precisely, b) in Ax2 and Ax4, there is a high significance as a function of speed, independently of running ability; however, c) in all amplitudes, the interaction between running ability and speed is not significant; finally, d) in sine and cosine functions, no significant differences are found as a function of both ability and speed.

VARIABLE	Ax1	Ax2	Ax3	Ax4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	0.01 **(p<0.01)**	p=NS	<0.0001 **(p<0.001)**
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Sinx1	Sinx2	Sinx3	Sinx4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosx1	Cosx2	Cosx3	Cosx4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

Table 18.9a. Amplitude coefficients and sine/cosine functions in the forward direction.

B. In both vertical and lateral directions, our results show that there are no differences between occasional, skilled and top runners no matter the investigated variables.

VARIABLE	Ay1 and Az1	Ay2 and Az2	Ay3 and Az3	Ay4 and Az4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Siny1 Sinz1	Siny2 Sinz2	Siny3 Sinz3	Siny4 Sinz4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

VARIABLE	Cosiny1 Cosinz1	Cosiny2 Cosinz2	Cosiny3 Cosinz3	Cosiny4 Cosinz4
Ability	p=NS	p=NS	p=NS	p=NS
Speed	p=NS	p=NS	p=NS	p=NS
Interaction Ability/Speed	p=NS	p=NS	p=NS	p=NS

Table 18.9b. Amplitude coefficients and sine/cosine functions in vertical and lateral directions.

3.2. Symmetry Index

As described in chapter 8 (par. 1), Symmetry Index has been calculated in each trial.

Single values of Symmetry Index (in runners) are contained in the enclosed CD (Second Study, Chapter 18, Symmetry Index and Symmetry Index statistical analysis).

3.2.1. Statistical analysis

Statistical analysis was performed by using each subject Symmetry Index value.

Results are presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The independent variable was progression speed (m/s). The chosen dependent variable was the Symmetry Index (SI) in each movement direction and running group.

A two-way ANOVA for related/unrelated measures (i.e. mixed design) was performed to compare the main effects of both speed and group. In addition, a post-*hoc* paired *t*-test with Bonferroni correction was used to detect the strength of the associations between single variables. The highest speed of 5.00 m/s has not been considered in this statistical analysis.

Therefore, firstly, results as a function of running speed will be presented and discussed; and secondly, comparisons among runner groups.

3.2.2. Graph legend

In each graph, the points represent mean values obtained by grouping the same running levels (OR *versus* SR *versus* TR) at different speeds.

Precisely, the average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 have been considered. For their single values, see Table 17.4, in chapter 17 (par. 2). Therefore, in the graphs below these speeds have been used. The lines represent the simple graphic amalgamation of all the data and the vertical bars the positive and negative standard deviations of the higher and lower speed curves (mean ± S.D.), respectively. Particularly:

- green indicates the forward direction;
- blue indicates the vertical direction;
- orange indicates the lateral direction.

As already described in chapter 8 (par. 4.1), SI ranges from 0.4 to 1.

3.2.3. Symmetry Index in occasional runners

All average Symmetry Index values are shown in Table 18.10a:

SYMMETRY INDEX in OCCASIONAL RUNNERS			
SPEED (m/s)	**Forward direction**	**Vertical direction**	**Lateral direction**
2.19	0.648 ± 0.067	0.909 ± 0.019	0.919 ± 0.031
2.74	0.679 ± 0.064	0.907 ± 0.024	0.900 ± 0.035
3.25	0.656 ± 0.136	0.879 ± 0.072	0.882 ± 0.047
3.90	0.640 ± 0.087	0.882 ± 0.031	0.845 ± 0.059
4.45	0.615 ± 0.096	0.847 ± 0.063	0.836 ± 0.065

Table 18.10a. Average Symmetry Indexes, occasional runners.

Therefore, in occasional runners (Figure 18.15), our results show that:

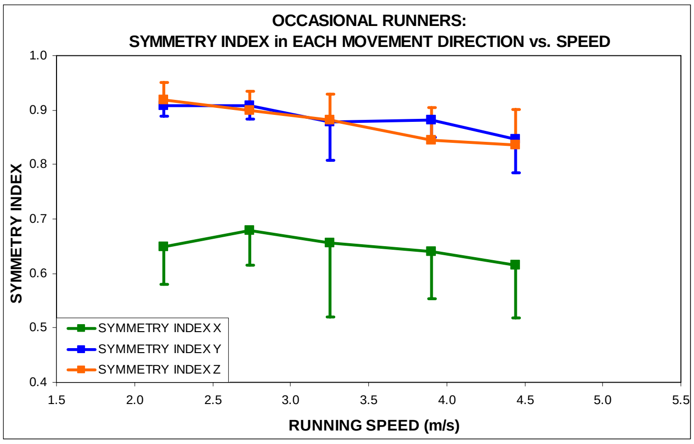


Figure 18.15. Symmetry Index along each movement direction, occasional runners.

- in the <u>forward direction</u>, SI increases with speed only from 2.19 to 2.74 m/s (from 0.648 ± 0.067 to 0.679 ± 0.064; p<0.01); however, it linearly decreases up to 4.45 m/s (from 0.656 ± 0.136 at 3.25 m/s to 0.615 ± 0.096 at 4.45 m/s, p<0.01). On average (0.648 ± 0.090, independently of speed), this is the least symmetrical direction;
- in the <u>vertical direction</u>, SI decreases with speed from 2.74 to 4.45 m/s (from 0.907 ± 0.024 to 0.847 ± 0.063, p<0.01). On average (0.885 ± 0.042), this is one of the most symmetrical direction;
- in the <u>lateral direction</u>, SI decreases with speed from 2.74 to 3.90 m/s (from 0.900 ± 0.035 to 0.845 ± 0.059, p<0.01). On average (0.876 ± 0.047), this is one of the most symmetrical direction.

Specific results of the statistical analysis (with relevance) are shown in Table 18.10b:

SPEED (m/s)	**MOVEMENT DIRECTION**		
	FORWARD	**VERTICAL**	**LATERAL**
2.19	p<0.01		
2.74		p<0.01	p<0.01
3.25	p<0.01		
3.90			
4.45			

Table 18.10b. SI as a function of speed in all movement directions, occasional runners.

3.2.4. Symmetry Index in skilled runners

All average Symmetry Index values are shown in Table 18.11a:

SYMMETRY INDEX in SKILLED RUNNERS			
SPEED (m/s)	**Forward direction**	**Vertical direction**	**Lateral direction**
2.23	0.670 ± 0.046	0.893 ± 0.027	0.906 ± 0.040
2.80	0.711 ± 0.048	0.892 ± 0.031	0.895 ± 0.019
3.34	0.735 ± 0.053	0.891 ± 0.030	0.885 ± 0.044
3.89	0.740 ± 0.042	0.884 ± 0.030	0.885 ± 0.042
4.45	0.755 ± 0.053	0.869 ± 0.029	0.884 ± 0.042
4.99	0.706 ± 0.075	0.864 ± 0.066	0.856 ± 0.060

Table 18.11a. Average Symmetry Indexes, skilled runners.

Therefore, in skilled runners (Figure 18.16), our results show that:

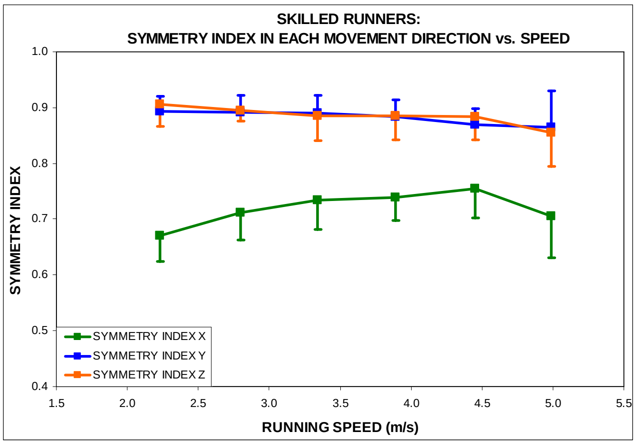


Figure 18.16. Symmetry Index along each movement direction, skilled runners.

- in the <u>forward direction</u>, SI increases with speed from 2.23 to 3.89 m/s (from 0.670 ± 0.046 to 0.740 ± 0.042, p<0.01); however, it decreases up to 4.99 m/s (from 0.755 ± 0.053 at 4.45 m/s to 0.706 ± 0.075 at 4.99 m/s, p<0.01). On average (0.720 ± 0.053, independently of speed), this is the least symmetrical direction;
- in the <u>vertical direction</u>, SI slightly decreases with speed from 3.34 to 4.99 m/s (from 0.884 ± 0.030 to 0.864 ± 0.066, p<0.05). On average (0.882 ± 0.036), this is one of the most symmetrical direction;

- in the lateral direction, SI slightly decreases with speed from 4.45 to 4.99 m/s (from 0.884 ± 0.042 to 0.856 ± 0.060, $p<0.05$). On average (0.885 ± 0.041), this is one of the most symmetrical direction.

Specific results of the statistical analysis (with relevance) are shown in Table 18.11b:

SPEED (m/s)	**MOVEMENT DIRECTION**		
	FORWARD	**VERTICAL**	**LATERAL**
2.23	$p<0.01$		
2.80			
3.34		$p<0.05$	
3.89			
4.45	$p<0.01$		$p<0.05$
4.99			

Table 18.11b. SI as a function of speed in all movement directions, skilled runners.

3.2.5. Symmetry Index in top runners

All average Symmetry Index values are shown in Table 18.12a:

SYMMETRY INDEX in TOP RUNNERS			
SPEED (m/s)	**Forward direction**	**Vertical direction**	**Lateral direction**
2.24	0.675 ± 0.046	0.867 ± 0.052	0.856 ± 0.037
2.66	0.709 ± 0.045	0.866 ± 0.054	0.832 ± 0.038
3.33	0.720 ± 0.027	0.864 ± 0.059	0.808 ± 0.055
3.89	0.745 ± 0.038	0.844 ± 0.047	0.796 ± 0.040
4.44	0.719 ± 0.068	0.822 ± 0.054	0.767 ± 0.052
5.00	0.772 ± 0.025	0.834 ± 0.055	0.791 ± 0.012

Table 18.12a. Average Symmetry Indexes, top runners.

Therefore, in top runners (Figure 18.17), our results show that:

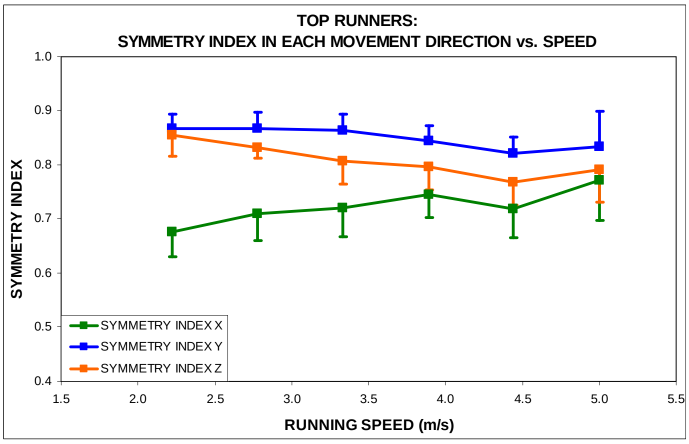


Figure 18.17. Symmetry Index along each movement direction, top runners.

- in the forward direction, SI highly decreases with speed from 3.89 to 4.44 m/s (from 0.745 ± 0.038 to 0.719 ± 0.068, $p<0.001$); however, it increases up to 5.00 m/s (from 0.719 ± 0.068 to 0.772 ± 0.025, $p<0.001$). On average (0.723 ± 0.042, independently of speed), this is the least symmetrical direction;
- in the vertical direction, SI slightly decreases with speed from 3.33 to 4.44 m/s (from 0.864 ± 0.059 to 0.822 ± 0.054, $p<0.05$); however, it increases up to 5.00 m/s (from 0.822 ± 0.054 to 0.834 ± 0.055, $p<0.01$). On average (0.849 ± 0.053), this is the most symmetrical direction;
- in the lateral direction, SI decreases with speed from 2.24 to 4.44 m/s (from 0.856 ± 0.037 to 0.767 ± 0.052, $p<0.01$); however, it increases up to 5.00 m/s (from 0.767 ± 0.052 to 0.791 ± 0.012, $p<0.01$). On average (0.808 ± 0.039), this is one of the most symmetrical direction.

Specific results of the statistical analysis (with relevance) are shown in Table 18.12b:

SPEED (m/s)	**MOVEMENT DIRECTION**		
	FORWARD	**VERTICAL**	**LATERAL**
2.24			$p<0.01$
2.66			
3.33		$p<0.01$ to 4.44 m/s $p<0.05$ to 5.00 m/s	
3.89	$p<0.001$		
4.44			
5.00			

Table 18.12b. SI as a function of speed in all movement directions, top runners.

3.3. Discussion

Observing all these results, we come to the conclusion that:

- Symmetry Index is slightly dependent on speed, in each runner group;
- independently of running level, right/left steps are more symmetrical in the vertical and lateral directions; however, they are less symmetrical in the forward direction;
- as expected, in all runners, the higher is the speed, the more asymmetrical is the corresponding movement of BCOM;
- finally, as expected, in the forward direction (Table 18.13), skilled and top runners are more symmetrical than occasional runners ($p<0.001$). However, in both vertical and lateral direction, top runners are more asymmetrical compared to the other groups ($p<0.01$). Indeed, in occasional and skilled runners no significant differences are found. This result does not agree with literature (Karamanidis et al., 2003).

SYMMETRY INDEX	**Forward direction**	**Vertical direction**	**Lateral direction**
OR *versus* SR	p<0.001 (independently of speed)	p=NS	p=NS
SR *versus* TR	p=NS	p<0.01	p<0.01
OR *versus* TR	p<0.01	p<0.01	p<0.01

Table 18.13. Single comparisons in Symmetry Index among runner groups.

Furthermore, in order a) to better understand and explain these results and b) to fully describe the symmetry of the BCOM, the mean overall Symmetry Index has been calculated, as well. For more details about this mathematical index, see chapter 8 (par. 6.1).

All mean overall Symmetry Index values are shown in Table 18.14, and they have been graphically represented in Figure 18.18 (occasional, skilled and top runners, respectively), as well.

SPEED (m/s)	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
2.22	0.825 ± 0.154	0.823 ± 0.132	0.799 ± 0.107
2.78	0.829 ± 0.130	0.833 ± 0.105	0.802 ± 0.083
3.33	0.806 ± 0.130	0.837 ± 0.088	0.797 ± 0.072
3.89	0.789 ± 0.130	0.836 ± 0.084	0.797 ± 0.049
4.44	0.766 ± 0.131	0.836 ± 0.071	0.769 ± 0.052
5.00	result no available	0.808 ± 0.089	0.799 ± 0.032

Table 18.14. Mean overall Symmetry Indexes, all runners.

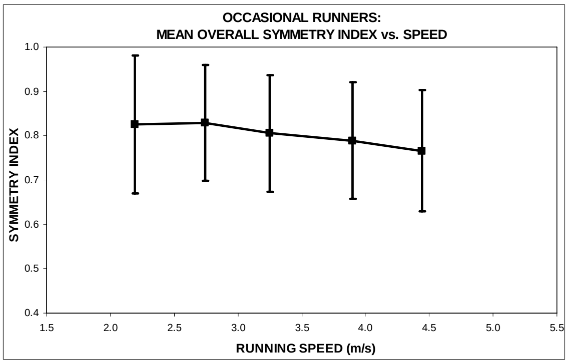


Figure 18.18a. Mean overall Symmetry Index, occasional runners.

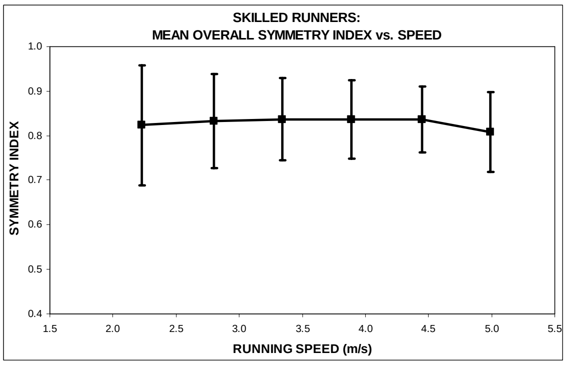


Figure 18.18b. Mean overall Symmetry Index, skilled runners.

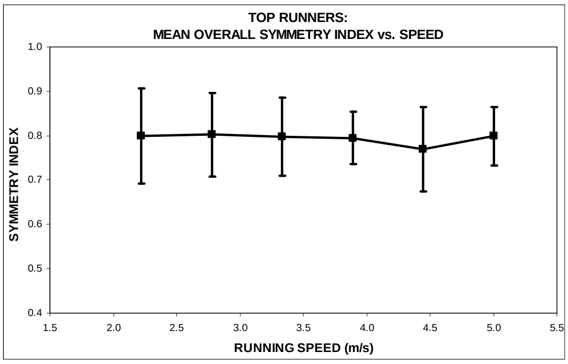


Figure 18.18c. Mean overall Symmetry Index, top runners.

Independently of movement direction, mean overall SI is slightly higher in skilled runners (0.829 ± 0.029: Figure 18.18b) in comparison to occasional (0.803 ± 0.137: Figure 18.18a) and top runners (0.794 ± 0.066: Figure 18.18c). This result seems a bit strange because it does not wholly confirm the initial hypothesis that ‘whether a runner is more trained, then his BCOM moves in a more symmetrical way’. Possible explanations could be found in:

- a quite high similarity in training level between all runners (see also chapter 19, par. 5) as the high standard deviations we obtained clearly show;
- some kinematic differences could occur after a more prolonged running time. Indeed, only few minutes of running (as proposed in our test protocol) could not be a time enough to determine evident discrepancies as a function of running ability. Therefore, we propose to maintain a single speed longer to better isolate and investigate kinematic adjustments in different trained runners.

4. POLAR GRAPHS

4.1. Introduction

In the following sections, the main results obtained by graphically representing both amplitude (A) and phase (φ) in a polar graph have been illustrated and discussed. To be precise, polar graphs have been drawn up following the same procedures described in chapter 9 (par. 3.3.3).

Indeed, average harmonic symmetrical and asymmetrical coefficients (with standard deviations) were used.

The graph legend discussed in chapter 9 (par. 3.1 and par. 3.3.3) has been used.

Clearly, it will be redundant to insert all single polar graphs for each runner group; therefore, we put in some of the most representative examples. Whereas a graphical example is not proposed, it means that it is quite similar to the other tested groups.

In detail, all radii, standard deviations and polar graphs (in all runners) are contained in the enclosed CD (Second Study, Chapter 18, Polar log graphs and radii variable).

4.2. Polar graphs in all runner groups

Skilled runners (running from 2.22 to 5.00 m/s) seem to have the most clear and regular graphical representations of symmetrical coefficients (x2, y2 and z1: left graph; x4, y4 and z3: right graph). This is an important result strongly depending on their major right/left symmetry (see par. 3 above). Therefore, the reference polar graphs below refer to this group.

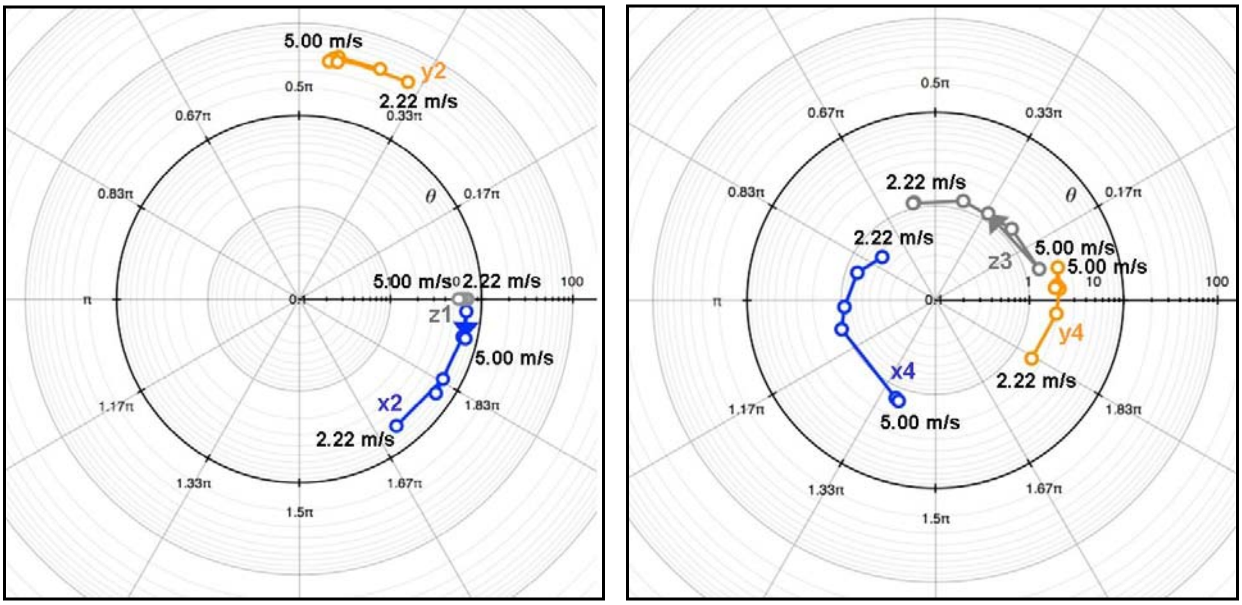


Figure 18.19. x2, y2 and z1 (on the left); x4, y4 and z3 (on the right) coefficients, skilled runners.

The qualitative analysis has shown that, compared to skilled runners, in occasional runners:

- x2 pattern is quite similar. Indeed, coefficients are close to 1.67π: this means that there has been a slightly downward shift;
- y2 pattern is quite similar, as well. Indeed, coefficients range from 0.33 to 0π meaning that there has been a slightly downward shift;
- z1 pattern is wholly similar to skilled runners;
- moreover, x4 shifts to an upward section: indeed, it ranges from 0.5 to 0.33π;
- y4 pattern is quite similar. However, the most evident difference is the opposite direction of the arrow;
- finally, z3 pattern is wholly similar to skilled runners.

Furthermore, compared both to skilled and occasional runners, in top runners:

- x2 pattern is quite similar to occasional runners. Indeed, coefficients are close to 1.67π: this highlights the slightly downward shift;
- y2 pattern is wholly similar to both skilled and occasional runners;
- however, z1 pattern has a slightly wider range;
- as shown in occasional runners, x4 shifts to an upward section ranging from 0.67 to 0π;

- y4 seems to have a relevant different pattern: indeed, it has been shifted to the opposite section;
- finally, z3 pattern is wholly similar to both skilled and occasional runners.

Occasional runners (running from 2.22 to 4.44 m/s) seem to have the most clear and regular graphical representations of asymmetrical coefficients (x1, y1 and z2: left graph; x3, y3 and z4: right graph). Therefore, the reference polar graphs below refer to this group.

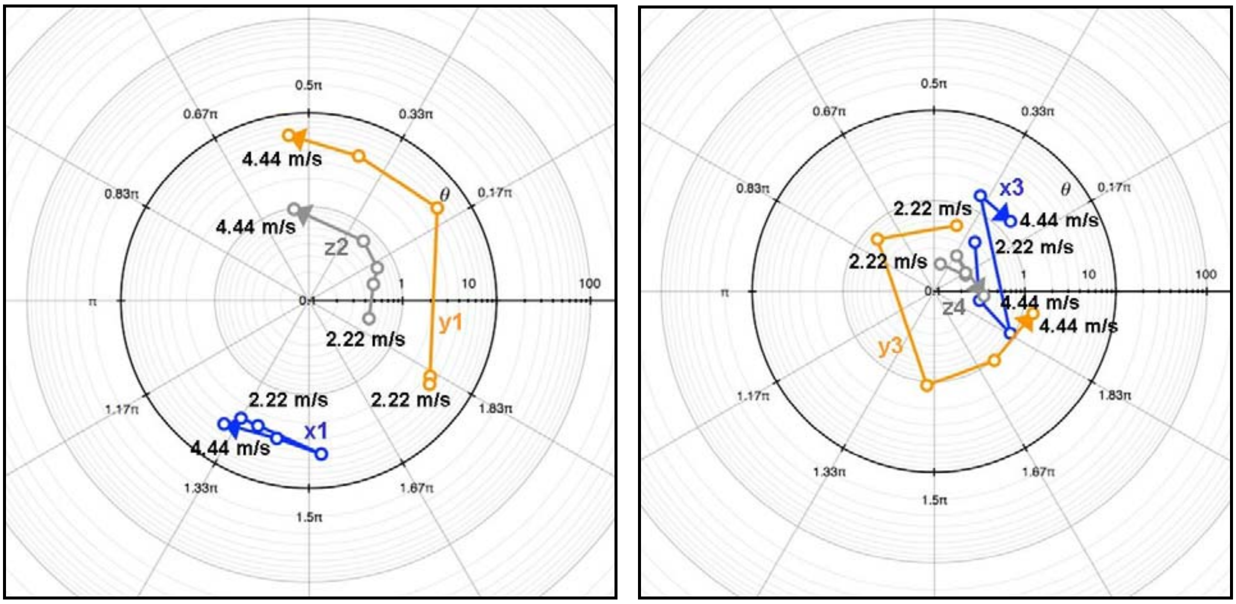


Figure 18.20. x1, y1 and z2 (on the left); x3, y3 and z4 (on the right) coefficients, occasional runners.

The qualitative analysis has shown that, compared to occasional runners, in skilled runners:

- x1 pattern is wholly similar;
- y1 and z2 seem to have an anomalous behaviour (Figure 18.21 below, left graph);
- moreover, x3 is wholly similar;
- y3 pattern is quite similar. However, the most evident difference is the opposite direction of the arrow;
- finally, z4 pattern is wholly similar to occasional runners.

Furthermore, compared both to skilled and occasional runners, in top runners, all coefficients seem to have a more regular and definite pattern (Figure 18.21 below, right graph).

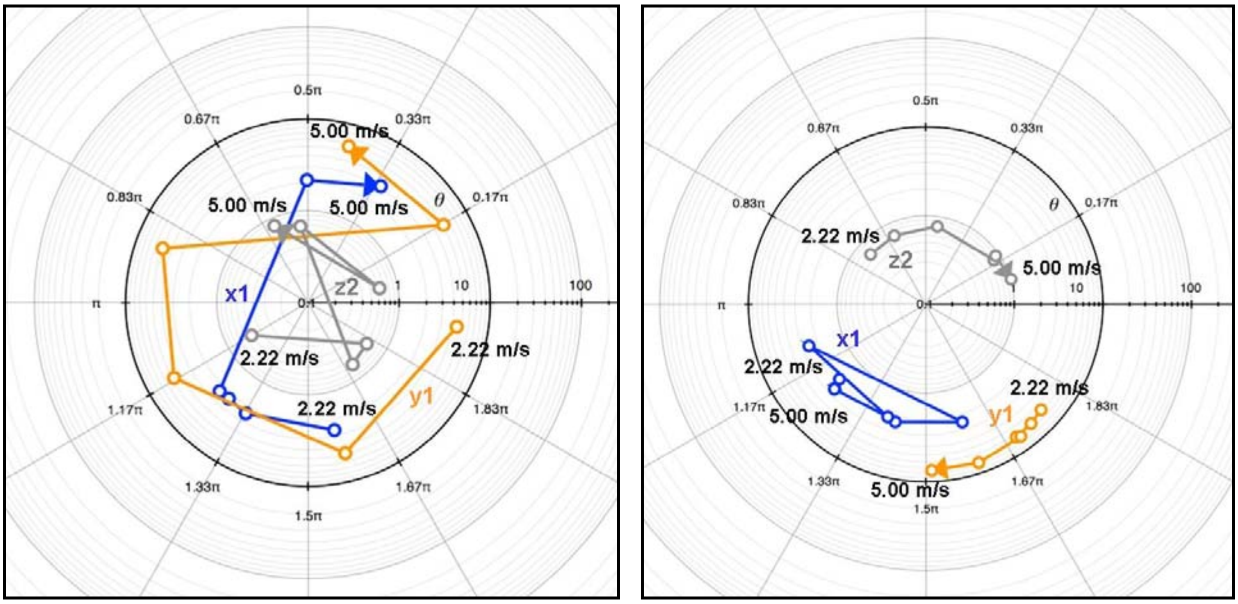


Figure 18.21. x1, y1 and z2, skilled (on the left) and top runners (on the right).

4.3. Discussion

We have tried to define an average fitting and behaviour of both amplitude and phase in a polar graph. Our results show that in symmetrical coefficients, there are only slight qualitative differences among groups. In other words, this means that if running speed increases, the pattern of each coefficient seems to change in a quite similar way independently of running ability.

However, the most evident differences have been found in the asymmetrical coefficients, especially in skilled and top runners. Because of the strong match in amplitudes, this is probably related to changeable phase patterns.

To verify this hypothesis, we suggest to exclude those subjects whose phases assume a value very different to runners of the same group. Thus, it implies a deeper observation of single data.

5. RUNNING ECONOMY

5.1. Introduction

As described in chapter 17 (par. 3), metabolic cost has been used to estimate running economy.

Single file derived from K4b^2 are contained in the enclosed CD (Second Study, Chapter 18, File K4).

Statistical analysis was performed by using each subject metabolic cost value.

Results are presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The independent variable was progression speed (m/s). The chosen dependent variable was the metabolic cost (C). Specifically, differences both among speed and group were assessed by using a one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test (with Bonferroni correction).

In each graph, the points represent mean values obtained by grouping the same running levels (OR *versus* SR *versus* TR) at different speeds. To be precise, the average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 have been considered (see chapter 17, par. 2). The lines represent the simple graphic amalgamation of all the data and the vertical bars the positive and negative standard deviations of the higher and lower speed curves (mean ± S.D.), respectively. To be precise:

- red corresponds to occasional runners' performance;
- green corresponds to skilled runners' performance;
- ski-blue corresponds to top runners' performance.

Finally, the other main physiological variables (i.e. ventilation, oxygen consumption, heart rate and respiratory exchange ratio; for more details, see also chapter 19, par. 3.3) are only graphically represented in the enclosed CD (Second Study, Chapter 18, Physiological variables). Indeed, they do not contribute in mainly characterizing running economy.

5.2. Results of our experiments

The average values of C (and relative S.D.) at each speed are shown in Table 18.15a

METABOLIC COST (J/(kg·m))			
SPEED (m/s)	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
2.22	4.93 ± 0.60	5.16 ± 0.73	4.83 ± 0.42
2.78	4.94 ± 0.47	4.93 ± 0.60	4.71 ± 0.22
3.33	4.78 ± 0.41	4.80 ± 0.57	4.55 ± 0.27
3.89	4.83 ± 0.52	4.69 ± 0.48	4.63 ± 0.34
4.44	4.58 ± 0.67	4.56 ± 0.38	4.70 ± 0.44
5.00	result no available	4.39 ± 0.36	4.71 ± 0.25

Table 18.15a. Metabolic cost at the different speeds during treadmill trials.

and graphically represented in Figure 18.22:

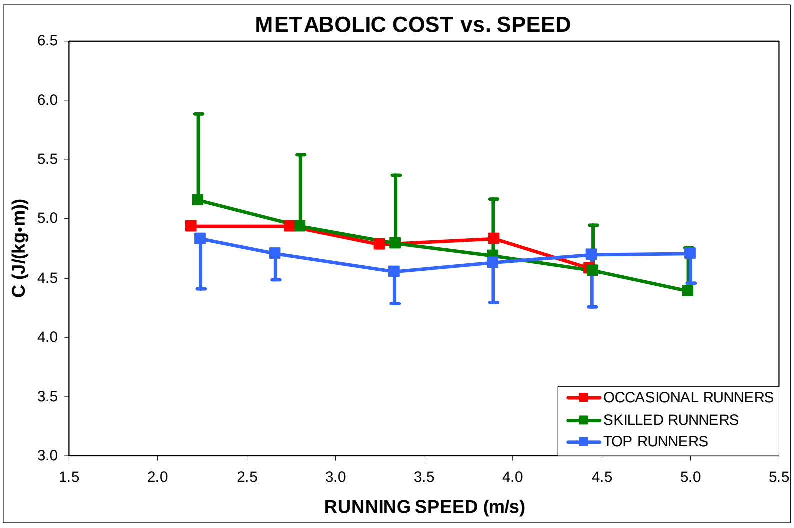


Figure 18.22. Metabolic cost *versus* running speed in treadmill trials.

As already widely demonstrated in literature (Margaria et al. 1963; di Prampero, 1985; Steudel, 1990; Martin et al., 1992; Guezennec et al., 1996; Dalleau et al., 1998; Malatesta et al., 2003), the energy cost values seem to be independent of speed, in each running level.

Furthermore, among all groups (Table 18.15b), there are no statistical significant differences:

RUNNERS	METABOLIC COST
OR *versus* SR	p=NS
SR *versus* TR	p=NS
OR *versus* TR	p=NS

Table 18.15b. Single comparisons in metabolic cost among runner groups.

5.3. Discussion

The lack of significant differences in various trained runners could be explained in:

- a quite high similarity in training level between all runners (Lees et al., 1994);
- a too much shorter running time. Indeed, some physiological differences could occur after a more prolonged running time so that only few minutes of running could not be a time enough to determine evident discrepancies as a function of running ability. Therefore, we propose to maintain a single speed longer to better isolate and investigate physiological adjustments in different trained runners. In such a way, for instance, metabolic cost could be recorded after 1 hour running.

Furthermore, in all movement directions, independently of running ability, the relationship between metabolic cost and Symmetry Index does not point out statistically significant results, as shown in Figure 18.23:

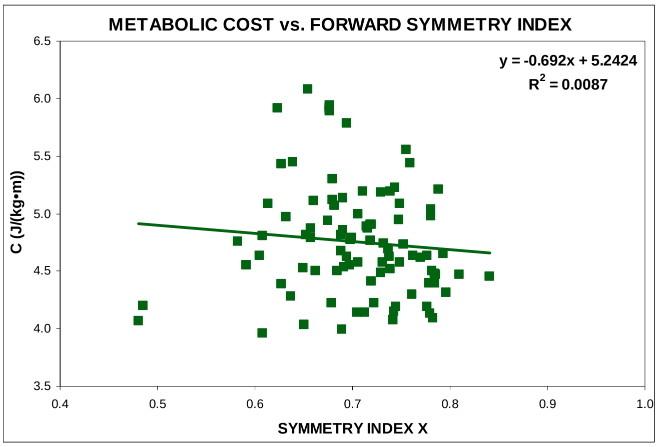


Figure 18.23a. Metabolic cost as a function of forward Symmetry Index, all runners.

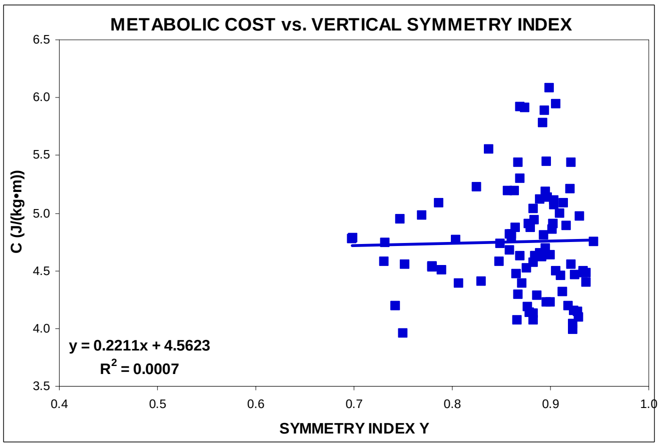


Figure 18.23b. Metabolic cost as a function of vertical Symmetry Index, all runners.

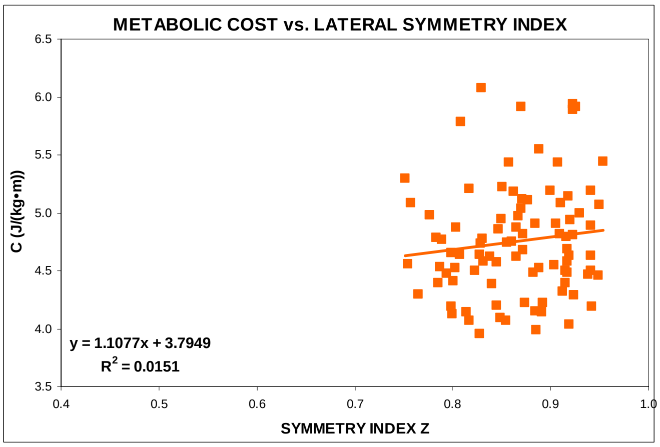


Figure 18.23c. Metabolic cost as a function of lateral Symmetry Index, all runners.

Particularly: a) in forward direction, n = 97, $R^2 = 0.0087$ and r = 0.0933; b) in vertical direction, n = 97, $R^2 = 0.0007$ and r = 0.0264; and c) in medial/lateral direction, n = 97, $R^2 = 0.0151$ and r = 0.1229.

6. CONCLUSION

The **2D analysis** of static anatomical symmetries (both leg and thigh segment) has shown that:

a) there are only slightly differences in right and left area among runners, independently of segment, area or percentage that have been investigated;

b) qualitatively, in some circumstances, top runners seem to have a more evident and increased area (right or left) in comparison to both occasional and skilled runners;

c) in general, this method seems to be not so adequate and valid to assess and test the main differences between right and left regions.

These results are probably related both to the application of a bi-dimensional approach, which can't be exhaustive and thorough, and to a high similarity among subjects.

Moreover, the **mathematical 3D analysis** of static anatomical symmetries has shown that:

d) there are no significant differences as a function of running ability, independently of anatomical regions;

e) importantly, there is a close match between the ankle and the knee area symmetries.

In addition, **Digital Locomotory Signature** has highlighted: 1) the more vertical displacement of the BCOM as a function of speed; 2) the presence of a right and left side asymmetry in the strides; and 3) the major left side asymmetry, probably related to the predominance of right-hand. These observations are independent of the subjects' level ability.

Right and left steps are more asymmetrical in occasional runners; in general, skilled runners are the most symmetrical subjects. Independently of running level, **Symmetry Index** is greater in vertical and lateral directions.

Symmetrical coefficients that have been plotted in a polar graph are quite similar among runners; however, slight differences could be appreciated in asymmetrical coefficients.

Finally, at all running speeds, there are no differences among groups.

Therefore, it seems that both a more symmetrical pattern of BCOM and a greater running economy are not related to a higher training level. Indeed, while these results are confirmed within dynamical functional symmetries, they are not within both static anatomical symmetries and running economy.

In any case, because our results do not wholly confirm literature data, it will be necessary to study a higher number of subjects with more defined different level training for further analysis.

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Chapter 19

TREADMILL *VERSUS* OVER-GROUND: BIOENERGETICS OF RUNNING

1. INTRODUCTION

A number of studies have compared bioenergetics and biomechanics of human running on surfaces with different stiffness (Nelson et al., 1972; Haisman et al., 1974; Andolf et al., 1976; McMahon et al., 1979; Frishberg, 1983; Pearce et al., 1983; Murray et al., 1985; Zamparo et al., 1992; Frangolias et al., 1995; Jones et al., 1996; Bosco et al., 1997; Alton et al., 1998; Lejeune et al., 1998; Wank et al., 1998; Sparrow, 2000; Kerdok et al., 2002; Meyer et al., 2003; Hall et al., 2004; Morin et al., 2005; Moritz et al., 2005; Nymark et al., 2005; Vuorimaa, 2005; Lee et al., 2007; Brughelli et al., 2008; Bowtell et al., 2009; Gordon et al., 2009; Terrier et al., 2009).

In this chapter, we will therefore explore the bioenergetics of human running on two different types of terrain: 1) **treadmill** and 2) **over-ground** (i.e. track).

Thus, it should be possible to specify the main similarities and/or differences between these two terrains. Before proceeding, it may be important to focus on the most important peculiarities between treadmill and over-ground locomotion that have been already discussed in literature.

2. TREADMILL LOCOMOTION *VERSUS* OVER-GROUND LOCOMOTION

2.1. Introduction

In recent years there has been a growing interest by athletes and coaches in physiological tests. Positively, this has focused attention on the validity of transferring information gained in the laboratory (i.e. treadmill) to the outdoor environment (i.e. track) (Jones et al., 1996).

The applicability of the treadmill is a constant topic of discussion among researchers and has resulted in wide investigation. Specifically, the treadmill offers a controlled and convenient environment for testing and training so that treadmills are commonly used in clinical and research settings for gait analysis or retraining (Schache et al., 2001; Wass et al., 2005). To ensure validity in locomotion research when using a treadmill, it is essential that the treadmill environment is as close to the over-ground environment as possible (Alton et al., 1998).

Nowadays, a lot of research has been developed to compare over-ground with treadmill locomotion, mostly a) to validate previous findings of treadmill locomotion or b) to justify the use of treadmill in future research.

The use of treadmills rests upon the assumption that reliable and valid measures of gait occur during treadmill locomotion, comparable to gait measures in over-ground one. In general, it has been known that treadmill locomotion may alter some aspects of neural control of a successive motor task, by imposing unusual operational modes on the nervous system (Zanetti et al., 2007).

In theory, treadmill walking is mechanically equivalent to over-ground walking. In reality, walking on a treadmill can initially be an unfamiliar experience (i.e. a new environment; Alton et al., 1998). Importantly, '*what happens in human treadmill running?*'

Literature focusing on the comparison between treadmill and over-ground locomotion (walking and running) states that the main variables investigated are kinematic, kinetic and cardio-respiratory patterns. Consequently, the next sections will briefly focus on these different factors.

2.2. Kinematic variables

2.2.1. Introduction

As regards kinematics, some studies found a higher cadence (steps *per* second) on the treadmill compared to the over-ground conditions (Murray et al., 1985; Milani et al., 1988; Lafortune et al., 1994; Stolze et al., 1997; Alton et al., 1998; Savelberg et al., 1998; White et al., 1998; Rodano et al., 2000; Schache et al., 2001; Kang et al., 2002; Bayat et al., 2005; Nymark et al., 2005; Brouwer et al., 2009; Carpinella et al., 2010). Particularly, stride length is decreased, stride rate increased and the period of non-support is less when running on a treadmill that when running over-ground (Elliot et al., 1976). Consequently, the stance and swing phases are shortened on the treadmill (Wank et al., 1998; Minetti et al., 2003; Wheat et al., 2005).

The duty factor appeared to be similar during treadmill walking (Warabi et al., 2005; Karamanidis et al., 2006) and during the higher running speeds (Nelson et al., 1972; Elliott et al., 1976; Stolze et al., 1997; Karamanidis et al., 2006).

Self-selected walking velocity was reported to be lower on the treadmill (Wank et al., 1998; Bayat et al., 2005; Brouwer et al., 2009), both with a metronome-enforced cadence (Arsenault et al., 1986) and a self-selected cadence (Vogt et al., 2002). Balance-related gait parameters such as step width and foot rotation angles increased during treadmill locomotion (Stolze et al., 1997).

Firstly, treadmill locomotion is associated with significant reductions in a) pelvis excursion (Schache et al., 2001; Vogt et al., 2002), b) locomotion variability and c) knee angles (Wheat et al., 2005), with significant improvements in d) local dynamic stability (Dingwell et al., 2001). In addition, analysis of the ankles reveals several interesting differences between over-ground and treadmill locomotion (Nigg et al., 1995).

Secondly, during the stance phase of walking, people changed the position of their centre of pressure: it moves more towards the front of their feet (Warabi et al., 2005).

Furthermore, at heel strike, the knee was less extended on the treadmill (Frishberg, 1983; Murray et al., 1985; Wheat et al., 2005). This movement was accompanied by a decreased flexion in the hip. In addition, at the end of the stance phase, the hip angle seemed to differ, although contrasting results have been reported (Alton et al., 1998; Vogt et al., 2002; Karamanidis et al., 2006). Due to changes in knee and hip angles in combination with the shortened strides, the body does not descend as low as during the double-support periods on a treadmill and over-ground walking (Wank et al., 1998; Bayat et al., 2005; Brouwer et al., 2009). As a result, the vertical excursion of the head (Murray et al., 1985) and the centre of mass (Nelson et al., 1972; Wank et al., 1998) are shorter on the treadmill. During treadmill locomotion, the centre of mass of the subject is not stationary, but moves with a constant velocity, the same as the treadmill belt but in the opposite direction (Warabi et al., 2005). Finally, a constant velocity is very important, since all accelerations and decelerations will not produce forces affecting the subject.

VARIABLE	TREADMILL	OVER-GROUND
Stance phase	<	>
Cadence	+ 7% in adults + 10% in children 6 - 7 years	<
Length stride cycle	<	>
Propulsion phase	>	<
Hip and knee flexion	> and < respectively	<
Pelvic oscillation	<	>

Table 19.1. Treadmill locomotion *versus* over-ground locomotion (some aspects).

It is important to remember that visual information influences treadmill locomotion and associated measures of stability (Milgrom et al., 2003; Eaves et al., 2008).

In the following paragraphs, some main aspects regarding kinematic variables that have been previously investigated in literature have been briefly proposed.

2.2.2. Literature review

In effect, the differences in the kinematics between treadmill and over-ground running could be divided into a) systematic- and b) subject- dependent components (Nigg et al., 1995). By comparing over-ground and treadmill ambulation to investigate possible differences in gait temporal variables and leg joint kinematics, it has been found significant differences among gender (i.e. in females, the maximum hip flexion angle was significantly different with greater flexion occurring on the treadmill; in males, significant differences were noted between the two conditions for cadence and maximum knee flexion angle with greater values in the treadmill walking) (Alton et al., 1998).

Furthermore, treadmill locomotion may (and in fact does appear to) reduce the variability associated with the stride interval time series (Dingwell et al., 2001; Wheat et al., 2005), maintaining the long range correlations intact (Jordan et al., 2006). Among the others, it has been demonstrated that the kinematic trajectories of treadmill gait were similar to those of over-ground gait (Riley et al., 2008).

More recently, the hypothesis that, contrary to over-ground walking, treadmill walking has effects on stance variables supporting the assumption that the imposed locomotion activity is more critical to stance control than natural walking has been verified (Zanetti et al., 2007). These results have showed that: a) treadmill locomotion produced an effect on body orientation in space during the subsequent stance trials; and b) this different orientation consisted in a forward inclination of the body, not accompanied by increased body sway for a few minutes.

The different components of the mechanical energy (potential, kinetic and rotational) of the segments considering the differences between walking on treadmill and over-ground were investigated, as well (Correa et al., 2000). It has been observed, on treadmill, a) a shorter stride length, b) a faster stride rate, c) a less range of motion in knee and ankle joints, d) a less variability in the horizontal and vertical velocity of the body centre of mass, and, finally, e) a reduction in the mechanical energy costs, especially at the trunk and lower limbs.

Yet, in order to determine the relationship between energy recovery and speed during treadmill walking, it has been satisfactorily demonstrated that, although a useful tool, walking on a treadmill may not be a true representation of ground walking. Therefore, it could not be the most effective way to research or rehabilitate gait (Collett et al., 2007). The participants formed two distinct groups: 1) those with normal BCOM energy recovery that was similar to ground walking; and 2) those with low BCOM energy recovery that was different to ground walking. Despite the low energy recovery values, both groups produced the expected U-shaped oxygen cost speed curve with no significant difference between groups.

Finally, subjects adjust the stiffness of their stance leg during treadmill running to accommodate differences in surface stiffness (McMahon et al., 1979; Kerdok et al., 2002; Cavagna et al., 2005; Morin et al., 2005; Brughelli et al., 2008; Grimmer et al., 2008). In such a way, a similar centre of mass movement has been maintained regardless of surface stiffness (Ferris et al., 1997; 1998; 1999; Derrick et al., 2000).

2.3. Kinetic variables

As far as kinetics is concerned, the normalized average EMG signal shows very similar patterns during over-ground and treadmill walking (Murray et al., 1985; Nymark et al., 2005; Brouwer et al.,

2009) and running (Wank et al., 1998). Very limited information is available about ground reaction forces and strains during treadmill gait. However, during treadmill locomotion, a more stable gait pattern is observed (i.e. an altered landing style and a lower variability) (Murray et al., 1985).

Finally, very few differences were found in temporal gait parameters or leg kinetics between treadmill and over-ground walking. Conversely, sagittal plane joint moments were found to be quite different at the knee and hip joints, but similar at the ankle. Moreover, differences in muscle activity were observed between the two walking modalities (particularly, in the tibialis anterior throughout stance, and in the hamstrings, vastus medialis and adductor longus during swing) (Lee et al., 2007).

2.4. Cardio-respiratory variables

As regards cardio-respiratory variables, the most evident difference found between over-ground and treadmill locomotion is a significantly lower oxygen consumption during treadmill running at higher velocities (Maksud et al., 1971; Jones et al., 1996; Carter et al., 2000; Kang et al., 2002; Meyer et al., 2003; Saunders et al., 2004a). However, it has been observed that between mean values for $V'O_2$ at level treadmill *versus* over-ground running and grade treadmill *versus* over-ground running no statistically significant differences have been found (Bassett et al., 1985).

2.5. Some peculiarities of treadmill locomotion

Empirical findings show a number of peculiarities in the treadmill locomotion pattern and physiological parameters. Some changes in the locomotion patterns may be caused by some fluctuations in: a) the belt speed (especially, during treadmill running) (van Ingen Schenau, 1980); b) the presence of handrails which may alter physiological and biomechanical parameters (Manfre et al., 1994; Brouwer et al., 2009); c) the limited belt size (Nigg et al., 1995); d) the firmness of the moving surface (Savelberg et al., 1998); and e) the mass of the subject (i.e. heavy subjects encounter greater ground reaction forces, resulting in an increased friction of the treadmill belt; Savelberg et al., 1998).

In addition, one more important difference (and peculiarity) exists between the treadmill and the over-ground situation: the lack of air resistance (Pugh, 1970; 1971; Davies, 1980; 1981; Jones et al., 1996). This aspect does not depend on the properties of a specific treadmill, but it is inherent to the treadmill use (Noakes et al., 1990; Weltman et al., 1990). A subject experiences no air resistance during treadmill locomotion; especially at high speeds, the influence becomes quite substantial. No air resistance during treadmill locomotion has several potential consequences in gait characteristics such as: 1) a lower propulsive force required to achieve an equal velocity compared to the over-ground situation (Pugh, 1971; Davies, 1980); 2) a change in body posture (Jones et al., 1996); and 3) a reduced convective heat loss (Jones et al., 1996).

Finally, as previously mentioning (see par. 2.2.1 above), the visual information as perceived by the subject is completely different in treadmill and over-ground locomotion (Rieser et al., 1995; Prokop et al., 1997; Dickstein et al., 2004; Eaves et al., 2008).

3. OUR COMPARISON BETWEEN TREADMILL AND OVER-GROUND RUNNING

In the following sections, we will present how we investigated the bioenergetics of different trained runners on two types of terrain: 1) **treadmill** and 2) **track**. As far as these running trials were concerned, only energetic (cardio-respiratory) results are available. In fact, both kinematics (in track) and kinetics (both treadmill and track) data were not recorded (see also chapter 16).

Singles' metabolic cost (both on treadmill and in track) and other physiological variables are contained in the enclosed CD (Second Study, Chapter 19, Metabolic cost).

3.1. Subjects

The bioenergetics of human running was measured in 14 runners (see also chapter 16) featuring varying running levels: 7 occasional runners (OR) and 7 skilled runners (SR).

The top runners (TR) did not perform locomotion in the track so that they could be not involved in this type of analysis.

3.2. Test protocol

Each subject was requested to run at 6 different speeds: from 2.22 to 5.00 m/s; step 0.56 m/s (see also chapter 16, par. 2). However, the highest speed of 5.00 m/s has been only graphically represented because occasional runners were not able to maintain it.

Each trial was carried out at the level gradient, contrarily to what stated in Jones et al. (1996). As previously described in chapter 16, the bioenergetics of running (i.e. physiological or cardio-respiratory variables) was obtained directly by the portable metabograph K4b^2 (Figure 19.1 above; see also chapter 15, par. 2).

Figure 19.1. K4b^2 during experiments on the treadmill (on the left) and in the track (on the right).

3.2.1. Treadmill tests

Firstly, tests were performed on treadmill (for its main characteristics, see also chapter 4, par. 2). Five minutes of basic routine was proposed (Zamparo et al., 2000; Minetti et al., 2001; Ardigò et al., 2005; Gottschall et al., 2005; Modica et al., 2005; Sawicki et al., 2008; Zamparo et al., 2008). The subject had to remain in a natural upright posture.

Each speed was then maintained for at least 5 minutes with a progressive incremental order.

Between each speed, a rest period of at least 5 minutes was proposed (for more details, see also chapter 16, par. 2).

3.2.2. Track tests

Secondly, tests were performed on the over-ground (in the track).

The same basic routine time has been proposed before starting the running trials. Speed was determined and controlled by an operator who cycled a bike sideways on the runner. As shown in the example in Table 19.2, the time of 100 metres was used to better check subject's speed.

SPEED (m/s)	Time 1	Time 2	Lap Time (1 lap = 400 mt)	Half Time		
Rest Activity	0	5’15”				
2.22	6’00”	11’53”	2’53” 5’53”	45”/100 mt	180”/lap	2 laps
2.78	14’00”	18’48”	2’24” 4’48”	36”/100 mt	144”/lap	2 laps
3.33	21’00”	24’56”	2’00” 3’56”	30.2”/100 mt	121”/lap	2 laps
3.89	27’30”	32’43”	1’40” 3’24” 5’13”	26”/100 mt	103”/lap	3 laps
4.44	36’00”	41’51”	1’30” 3’00” 4’26” 5’51”	22.5”/100 mt	90”/lap	4 laps
5.00	46’30”	49’02”	1’16” 2’32”	20”/100 mt	80”/lap	4 laps

Table 19.2. An example of the speed’s control procedure.

To be more precise:

- Time 1 refers to the beginning of each running speed (minute);
- Time 2 refers to the ending of the same speed (minute);
- Lap Time refers to each single time to complete one lap (minute);
- Half Time refers to the time requested to run 100 metres at each speed.

All runners (both occasional and skilled) were able to respect these interval speeds.

As previously described in chapter 16, both on treadmill and in the track, only five skilled runners were able to complete all running protocols (up to 5.00 m/s). Other subjects (both occasional and skilled runners) stopped at the speed of 4.44 m/s.

Treadmill and track running tests were performed in the Biomechanics Laboratory and on a track, respectively, at the Faculty of Exercise and Sport Science at Verona University.

3.3. Main physiological parameters in bioenergetics of running

3.3.1. Definition of single variables

As widely demonstrated in literature, they are:

- **ventilation** (V’E): the rate at which gas enters or leaves the lung (Ventilation in Physiology - Wikipedia, the free encyclopedia, 2009), depending on both nervous and biochemical stimuli. During the basal activity, minute ventilation is ≈ 7.5 L/min; in running activity, it is ≈ 20-25 L/min (Zamparo et al., 2000; McArdle et al., 2001; Hoffman, 2005);

- **oxygen consumption** (or uptake, V’O_2): the rate at which oxygen is used by a tissue (Oxygen consumption in Physiology - Wikipedia, the free encyclopedia, 2009) or the rate at which oxygen enters the blood from alveolar gas, equal in steady state to the consumption of energy by tissue metabolism throughout the body (di Prampero, 1985; Sparrow, 2000; Kang et al., 2002; Raynor et al., 2002; Jørgensen et al., 2009);
- **heart rate** (HR): the average resting heartbeat per minute of the heart (Finkenzeller et al., 2009; Heart rate in Physiology - Wikipedia, the free encyclopedia, 2009; Huovinen et al., 2009);
- **respiratory equivalent** (RE): the ratio between ventilation (V’E) and oxygen consumption (V’O_2). In a young healthy male, it is usually 25:1. Precisely, it constitutes a fundamental parameter in defining the anaerobic threshold;
- **respiratory exchange ratio** (RER). In one breath, a healthy person normally breathes in more molecules of oxygen than he/she breathes out molecules of carbon dioxide: the ratio between these CO_2/O_2 is the respiratory exchange ratio (Mahaudens et al., 2008; 2009). Measuring this ratio can be used for estimating the **respiratory quotient** (RQ), an indicator of which fuel (carbohydrate or fat) is being metabolized to supply the body with energy (Respiratory exchange ratio in Physiology - Wikipedia, the free encyclopedia, 2009);
- **metabolic cost** (C): the metabolic energy spent to transport the body mass of a subject (Zamparo et al., 2000) or the amount of energy consumed *per* unit of distance (Slawinski et al., 2004). For this variable, see also chapter 17.

For other important characteristics and properties of these physiological variables, we suggest looking through specific tracts on Physiology (i.e. Margaria, 1938; Astrand et al., 1977; Dal Monte, 1983; di Prampero, 1985; Sjodin et al., 1985; Brisswalter et al., 1994; Taylor, 1994; Frangolias et al., 1995; di Prampero, 1997; Cerretelli, 2001; McArdle et al., 2001; Chen et al., 2009).

3.3.2. Calculation and statistical analysis of single variables

In both treadmill and track experiments, each parameter has been elaborated and calculated as: a) in the rest activity, the mean value of the last two minutes of single recordings; and b) in each running speed, the mean value of the last two minutes of single recordings.

In each testing condition (see also chapter 16, par. 2.2), average values were obtained (see also chapter 17, par. 3.5). They will be widely discussed in the following sections.

Moreover, statistical analysis was performed by using each physiological variable value. Precisely, results are presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The independent variable was progression speed (m/s). The chosen dependent variables were the ventilation (V’E), the oxygen consumption (V’O$_2$), the respiratory equivalent (RE), the heart rate (HR), the respiratory exchange ratio (RER) and the metabolic cost (C).

Differences between treadmill and track (in occasional and skilled runners) were assessed by using paired *t*-tests.

4. RESULTS IN TREADMILL AND OVER-GROUND RUNNING’S COMPARISON

4.1. Introduction: single values

Before analysing the pattern of each variable, in tables below (Table 19.3a, 19.3b, 19.3c, 19.3d and 19.3e), all their values are shown in both treadmill and track.

The distinction in the two runner groups was respected.

SPEED (m/s)	V’E (ml/min) on TREADMILL	V’E (ml/min) in the TRACK	V’E (ml/min) on TREADMILL	V’E (ml/min) in the TRACK
	OCCASIONAL RUNNERS		SKILLED RUNNERS	
Rest Activity	11.40 ± 2.09	10.08 ± 1.46	10.30 ± 1.62	10.88 ± 1.47
2.22	55.34 ± 8.24	48.10 ± 6.20	55.03 ± 14.65	50.35 ± 7.85
2.78	68.49 ± 11.46	60.28 ± 5.81	67.67 ± 15.37	61.77 ± 6.67
3.33	82.82 ± 11.23	76.87 ± 6.92	82.95 ± 17.40	73.37 ± 10.69
3.89	112.19 ± 22.84	101.41 ± 11.13	99.06 ± 21.12	86.27 ± 10.82
4.44	139.47 ± 23.59	131.99 ± 19.91	127.06 ± 22.22	118.37 ± 13.04
5.00	result no available		145.13 ± 22.29	148.33 ± 16.06

Table 19.3a. V’E on treadmill and in the track, occasional and skilled runners.

SPEED (m/s)	V’O$_2$ (ml/kg·min) on TREADMILL	V’O$_2$ (ml/kg·min) in the TRACK	V’O$_2$ (ml/kg·min) on TREADMILL	V’O$_2$ (ml/kg·min) in the TRACK
	OCCASIONAL RUNNERS		SKILLED RUNNERS	
Rest Activity	5.23 ± 0.93	3.80 ± 0.73	4.64 ± 0.60	4.26 ± 1.08
2.22	36.83 ± 4.21	32.47 ± 3.28	37.53 ± 4.89	33.13 ± 2.59
2.78	44.74 ± 4.37	39.44 ± 2.30	43.99 ± 5.10	39.57 ± 2.73
3.33	51.09 ± 4.61	48.52 ± 2.42	50.53 ± 5.68	45.98 ± 2.56
3.89	59.29 ± 5.97	55.73 ± 4.01	57.02 ± 5.50	52.57 ± 3.37
4.44	63.84 ± 8.48	62.81 ± 11.12	62.84 ± 4.58	59.79 ± 4.89
5.00	result no available		63.53 ± 4.72	64.26 ± 3.41

Table 19.3b. V’O$_2$ on treadmill and in the track, occasional and skilled runners.

SPEED (m/s)	RE on TREADMILL	RE in the TRACK	RE on TREADMILL	RE in the TRACK
	OCCASIONAL RUNNERS		SKILLED RUNNERS	
2.22	21.29 ± 1.54	21.26 ± 2.07	21.52 ± 2.41	22.55 ± 1.87
2.78	21.76 ± 2.23	21.98 ± 2.17	22.74 ± 3.07	23.28 ± 2.05
3.33	22.92 ± 1.87	22.77 ± 1.95	24.36 ± 3.53	23.73 ± 2.54
3.89	26.67 ± 3.91	26.14 ± 2.34	25.76 ± 4.07	24.79 ± 2.88
4.44	31.16 ± 5.43	30.81 ± 2.41	30.09 ± 4.22	29.95 ± 3.40
5.00	result no available		31.93 ± 2.66	34.44 ± 3.00

Table 19.3c. RE on treadmill and in the track, occasional and skilled runners.

SPEED (m/s)	**HR (bpm) on TREADMILL**	**HR (bpm) in the TRACK**	**HR (bpm) on TREADMILL**	**HR (bpm) in the TRACK**
	OCCASIONAL RUNNERS		**SKILLED RUNNERS**	
Rest Activity	87.49 ± 11.58	84.67 ± 14.18	72.66 ± 11.34	75.02 ± 8.45
2.22	134.57 ± 7.32	127.62 ± 12.94	124.41 ± 11.64	116.59 ± 8.96
2.78	150.53 ± 7.63	145.46 ± 13.60	138.99 ± 11.23	130.22 ± 11.46
3.33	166.67 ± 9.12	162.16 ± 11.96	152.75 ± 11.68	145.71 ± 8.63
3.89	179.01 ± 11.07	175.72 ± 11.24	166.93 ± 11.41	160.56 ± 8.83
4.44	186.31 ± 7.59	180.32 ± 6.75	178.16 ± 7.56	177.21 ± 10.64
5.00	result no available		183.57 ± 10.50	184.32 ± 10.16

Table 19.3d. HR on treadmill and in the track, occasional and skilled runners.

SPEED (m/s)	**RER on TREADMILL**	**RER in the TRACK**	**RER on TREADMILL**	**RER in the TRACK**
	OCCASIONAL RUNNERS		**SKILLED RUNNERS**	
Rest Activity	0.82 ± 0.05	0.83 ± 0.12	0.79 ± 0.04	0.77 ± 0.03
2.22	0.86 ± 0.05	0.85 ± 0.06	0.87 ± 0.07	0.83 ± 0.02
2.78	0.88 ± 0.02	0.88 ± 0.07	0.88 ± 0.05	0.88 ± 0.02
3.33	0.89 ± 0.05	0.91 ± 0.07	0.89 ± 0.07	0.90 ± 0.05
3.89	0.96 ± 0.07	0.98 ± 0.10	0.94 ± 0.05	0.91 ± 0.06
4.44	1.03 ± 0.08	1.03 ± 0.08	0.99 ± 0.06	0.98 ± 0.07
5.00	result no available		1.04 ± 0.05	1.09 ± 0.04

Table 19.3e. RER on treadmill and in the track, occasional and skilled runners.

In the following sections, we will isolate and define the specific pattern of each variable in order to identify the similarities or differences between occasional and skilled runners, both in treadmill and over-ground running.

4.2. Graph legend

In each graph, the points represent mean values obtained by grouping the same running levels (OR *versus* SR) at the different speeds; the lines the simple graphic amalgamation of all the data; and the vertical bars the positive and negative standard deviations of the higher and lower speed curves (mean ± S.D.), respectively.

Precisely, the average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 have been considered (see Table 17.4, in chapter 17, par. 2). However, in order to avoid confusions, in the tables above the main speeds (2.22, 2.78, 3.33, 3.89, 4.44 and 5.00 m/s) are reported. Moreover:

- red corresponds to treadmill performance (occasional and skilled runners);
- green corresponds to over-ground (track) performance (occasional and skilled runners).

4.3. Ventilation

Our results show that:

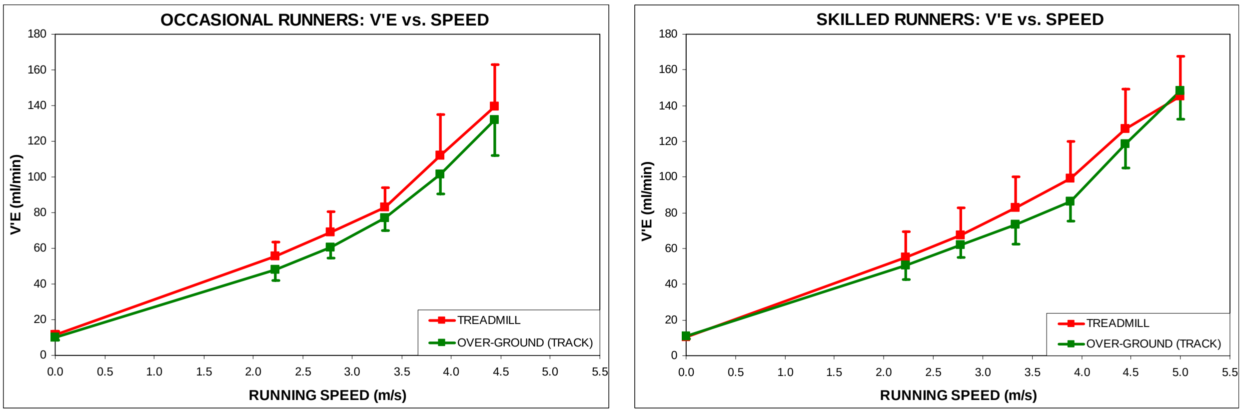

Figure 19.2. Ventilation as a function of running speed,
occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- V'E highly increases with speed, independently of both running levels and surfaces (Table 19.4: $p<0.001$). Precisely, on treadmill, in occasional runners: n=7, $R^2 = 0.9369$, r = 0.9679; in skilled runners: n = 7, $R^2 = 0.9699$, r = 0.9848; in the track, in OR: $R^2 = 0.9221$, r = 0.9603; in SR: $R^2 = 0.9183$, r = 0.9583;

SUBJECTS	**VENTILATION (V'E)**
OCCASIONAL RUNNERS	p<0.001 comparing each speed
SKILLED RUNNERS	p<0.001 comparing each speed

Table 19.4. V'E as a function of running speed.

- importantly, on average, our values of V'E are very similar to those reported in literature (Margaria, 1938; Boje, 1944; Dill, 1965; Pugh, 1970; Maksud et al., 1971; di Prampero, 1985; Brandon et al., 1992; Pate et al., 1992; Brisswalter et al., 1994; Berry et al., 1996; Guezennec et al., 1996; Klein et al., 1997; Wood, 1999; Kerdok et al., 2002; Avogadro et al., 2003; Meyer et al., 2003; Malison et al., 2004; Saunders et al., 2004b; Zamparo et al., 2008; Parvataneni et al., 2009);
- at the higher speeds of 3.89 and 4.44 m/s, the qualitative analysis has shown that V'E is slightly higher in occasional runners than in skilled, similarly on treadmill and in the track running, according to Kyrolainen et al. (1995);
- however, the statistical analysis has shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all the investigated speeds.

4.4. Oxygen consumption

Our results show that:

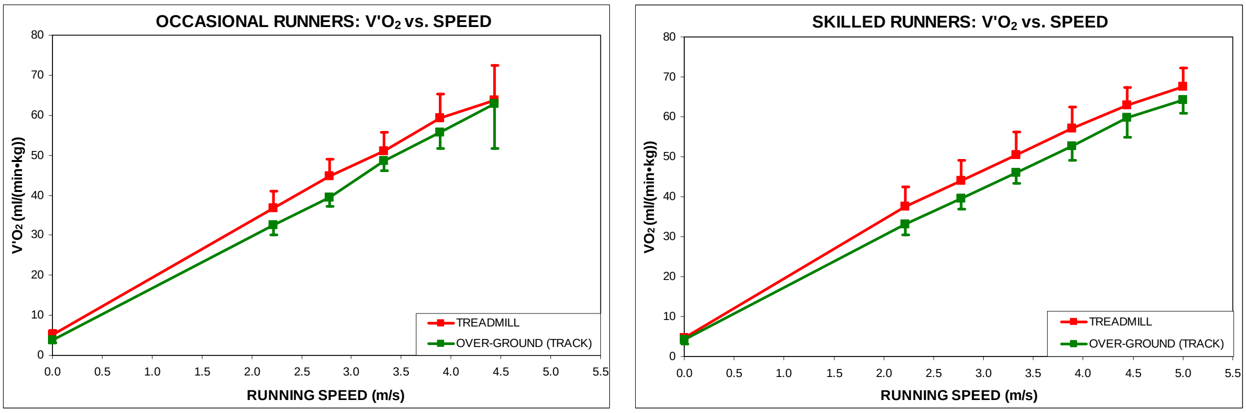

Figure 19.3. Oxygen consumption as a function of running speed, occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- V'O$_2$ highly increases with speed (Kang et al., 2002), independently of both running levels and surfaces (Table 19.5: $p<0.001$). Precisely, on treadmill, in OR: $R^2 = 0.9951$, $r = 0.9975$; in SR: $R^2 = 0.9969$, $r = 0.9984$; in the track, in OR: $R^2 = 0.9990$, $r = 0.9995$; in SR: $R^2 = 0.9977$, $r = 0.9988$;

SUBJECTS	OXYGEN CONSUMPTION (V'O$_2$)
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed

Table 19.5. V'O$_2$ as a function of running speed.

- importantly, on average, our values of V'O$_2$ are very similar to those reported in literature (Knuttgen, 1961; McMiken et al., 1976; Hagan et al., 1980; Daniels, 1985; di Prampero, 1985; Kaneko, 1990; Steudel, 1990; Brandon et al., 1992; Guezennec et al., 1996; Jones et al., 1996; Klein et al., 1997; Pereira et al., 1997; Wood, 1999; Carter et al., 2000; Kyrolainen et al., 2001; Raynor et al., 2002; Avogadro et al., 2003; Caputo et al., 2003; Meyer et al., 2003; Millet et al., 2003; Saunders et al., 2004a; 2004b; Rotstein et al., 2005; James et al., 2007; Marcovic et al., 2007; Zamparo et al., 2008; Fletcher et al., 2009; Jørgensen et al., 2009);
- the qualitative analysis has shown that V'O$_2$ is slightly lower in the track, independently of runners;
- however, the statistical analysis has shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all the investigated speeds.

4.5. Respiratory equivalent

Our results show that:

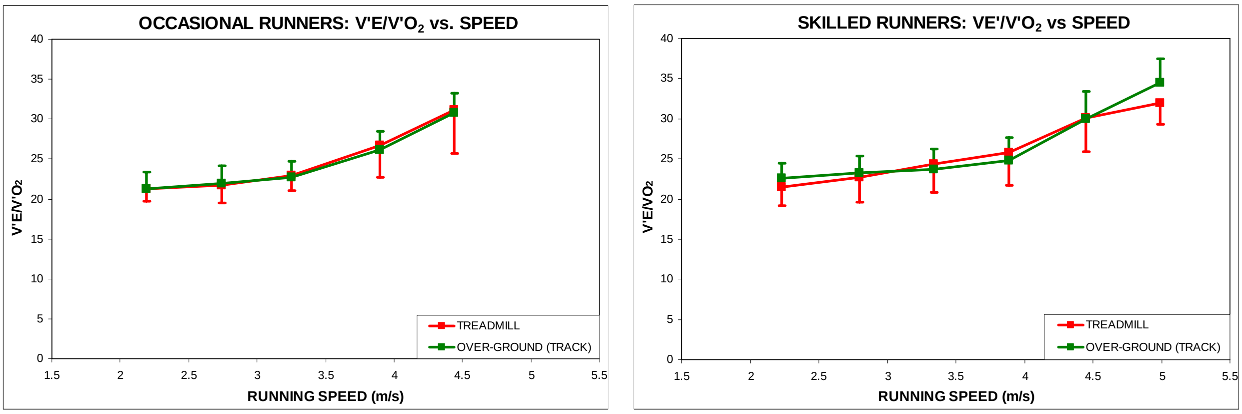

Figure 19.4. Respiratory equivalent as a function of running speed, occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- RE highly increases with speed, independently of both running levels and surfaces (Table 19.6: $p<0.001$). Precisely, on treadmill, in OR: $R^2 = 0.8958$, $r = 0.9465$; in SR: $R^2 = 0.9542$, $r = 0.9768$; in the track, in OR: $R^2 = 0.8673$, $r = 0.9312$; in SR: $R^2 = 0.9831$, $r = 0.9111$;

SUBJECTS	RESPIRATORY EQUIVALENT (RE)
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed

Table 19.6. RE as a function of running speed.

- importantly, the anaerobic threshold (di Prampero, 1985) seems to occur at the same point, independently of running levels and surfaces (Allen et al., 1985; Zacharogiannis et al., 1993; Frangolias et al., 1995; Avogadro et al., 2003; Skof et al., 2006);
- both the qualitative and the statistical analysis have shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all the investigated speeds.

4.6. Heart rate

Our results show that:

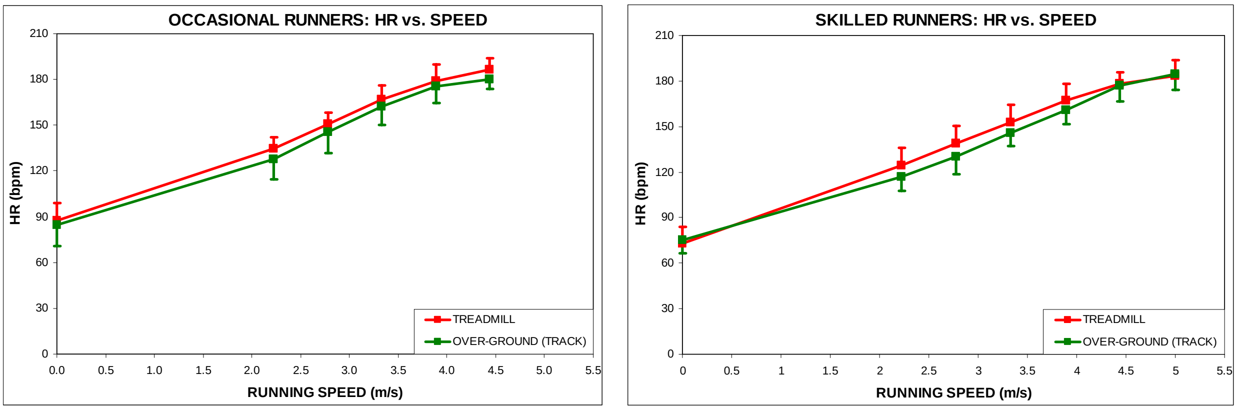

Figure 19.5. Heart rate as a function of running speed, occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- HR highly increases with speed, independently of both running levels and surfaces (Knuttgen, 1961; Londeree et al., 1982; Brisswalter et al., 1994; Jones et al., 1996; Hiilloskorpi et al., 1999; Kang et al., 2002; Meyer et al., 2003; Niskanen et al., 2004; Zamparo et al., 2008; Huovinen et al., 2009; Montgomery et al., 2009; Sibley et al., 2009) (Table 19.7: $p<0.001$). Precisely, on treadmill, in OR: R^2 = 0.9805, r = 0.9902; in SR: R^2 = 0.9831, r = 0.9915; in the track, in OR: R^2 = 0.8740, r = 0.9349; in SR: R^2 = 0.9901, r = 0.9950;

SUBJECTS	HEART RATE (HR)
OCCASIONAL RUNNERS	p<0.001 comparing each speed
SKILLED RUNNERS	p<0.001 comparing each speed

Table 19.7. HR as a function of running speed.

- on average, in rest activity, our heart rate values are slightly higher than those reported in literature (especially in occasional runners) (Margaria, 1938; Knuttgen, 1961; Pugh, 1970; Maksud et al., 1971; Murray et al., 1985; di Prampero, 1985; Pate et al., 1992; Guezennec et al., 1996; Klein et al., 1997; Wood, 1999; Kerdok et al., 2002; Malison et al., 2004; Saunders et al., 2004b; Rotstein et al., 2005; Richards, 2008; Zamparo et al., 2008; Parvataneni et al., 2009; Ruckstuhl et al., 2009);
- both in occasional and skilled runners, HR is never higher than 180 bpm (McArdle et al., 2001), suggesting that all subjects were able to complete all the test protocol;
- the qualitative analysis has shown that HR is slightly higher in occasional runners at all speeds (Bailey et al., 1991; Kyrolainen et al., 1995). However, the statistical analysis has shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all speeds. Indeed, comparing treadmill to track, independently

of running level (n = 63) and speed, R^2 = 0.8504 and r = 0.9922. This means that no significant differences exist among these terrains.

4.7. Oxygen consumption *versus* heart rate

In treadmill locomotion, it has been demonstrated that there is a linear relation between $V'O_2$ and HR through a wide range of speeds (Corry et al., 1996; Kang et al., 2002; Bar-Haim et al., 2008; Jørgensen et al., 2009). However, as widely shown, the limitations to the use of HR as an indicator of energy expenditure include:

1. the slope of the linear relationship between $V'O_2$/HR varies among individuals with different aerobic capacities (Berg et al., 1970; McGregor et al., 2009);
2. within a single individual at different times, the relation changes with the effect of training or de-training (Berg et al., 1970).

In spite such limitations, we have decided to investigate whether the relationship between these two variables could be significant among the two types of terrain. An eventual significant relationship could be related to an instrumentation error. Positively, its absence states that this hypothesis has to be rejected. Indeed, precisely, our results show that:

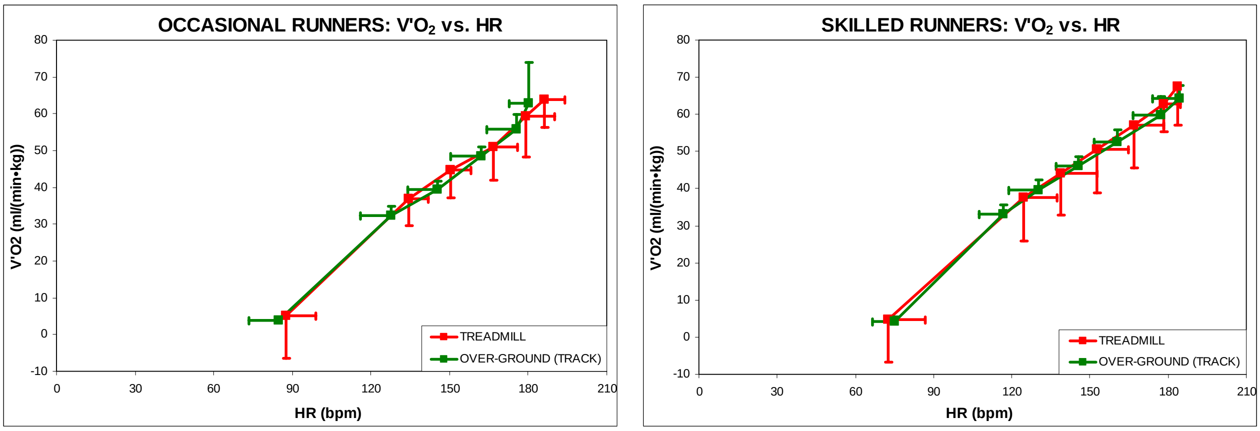

Figure 19.6. Oxygen consumption as a function of heart rate, occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- the relationship between oxygen consumption and heart rate highly increases with speed, independently of both running levels and surfaces (Table 19.8: $p<0.001$). Precisely, on <u>treadmill</u>, in OR: R^2 = 0.9934, r = 0.9967; in SR: R^2 = 0.9942, r = 0.9971; in the <u>track</u>, in OR: R^2 = 0.9908, r = 0.9954; in SR: R^2 = 0.9848, r = 0.9924;

SUBJECTS	OXYGEN CONSUMPTION ($V'O_2$) vs. HEART RATE (HR)
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed

Table 19.8. $V'O_2$ as a function of heart rate.

- the qualitative analysis has shown that, at the higher running speeds (3.89 and 4.44 m/s), this relationship is smaller in the track;
- however, the statistical analysis has shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all the investigated speeds.

4.8. Respiratory exchange ratio

Our results show that:

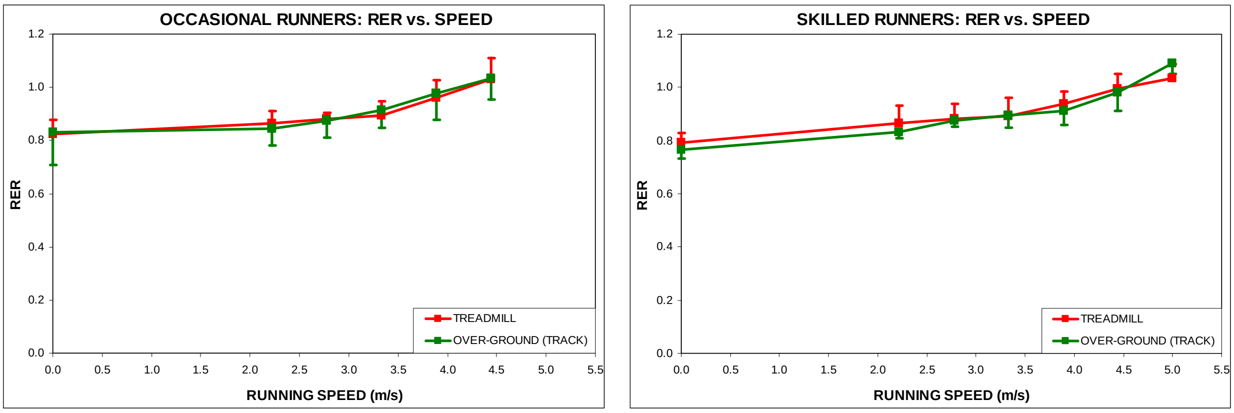

Figure 19.7. Respiratory exchange ratio as a function of running speed, occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

- as expected, RER is never higher than 1.1 (McArdle et al., 2001; Doke et al., 2004; Malison et al., 2004; Saunders et al., 2004b), suggesting that all subjects were able to complete the test protocol;
- RER highly increases with speed, independently of both running levels and surfaces (Table 19.9: $p<0.001$). Precisely, on treadmill, in OR: $R^2 = 0.8991$, $r = 0.9482$; in SR: $R^2 = 0.9939$, $r = 0.9695$; in the track, in OR: $R^2 = 0.9770$, $r = 0.9884$; in SR: $R^2 = 0.8903$, $r = 0.9436$;

SUBJECTS	RESPIRATORY EXCHANGE RATIO (RER)
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed

Table 19.9. RER as a function of running speed.

- both the qualitative and the statistical analysis have shown that there are no significant differences between OR and SR, in both treadmill and track: graphs are similar at all the investigated speeds.

4.9. Metabolic cost

All values of C in occasional and skilled runners on the two types of terrain are shown in the table below:

SUBJECTS	SPEED (m/s)	TREADMILL	TRACK
		C (J/(kg·m))	C (J/(kg·m))
OCCASIONAL RUNNERS	**2.22**	4.93 ± 0.60	4.41 ± 0.24
	2.78	4.94 ± 0.47	4.40 ± 0.29
	3.33	4.78 ± 0.41	4.62 ± 0.20
	3.89	4.83 ± 0.52	4.60 ± 0.33
	4.44	4.58 ± 0.67	4.52 ± 0.87
	5.00	result no available	
SKILLED RUNNERS	**2.22**	5.16 ± 0.73	4.53 ± 0.39
	2.78	4.93 ± 0.60	4.43 ±0.31
	3.33	4.80 ± 0.57	4.36 ± 0.32
	3.89	4.69 ± 0.48	4.31 ± 0.33
	4.44	4.56 ± 0.38	4.33 ± 0.43
	5.00	4.39 ± 0.36	4.17 ± 0.28

Table 19.10. C during treadmill and track running tests, occasional and skilled runners.

As expected, our results show that, both in OR and SR, metabolic costs are independent of speed in both terrains (Margaria, 1938; Conley et al., 1980; Burdett et al., 1983; Pearce et al., 1983; Carrier, 1984; di Prampero, 1985; Williams, 1990; Brisswalter et al., 1994; 1996; Candau et al., 1998; Raynor et al., 2002; Zamparo et al., 2008; Steudel-Numbers et al., 2009).

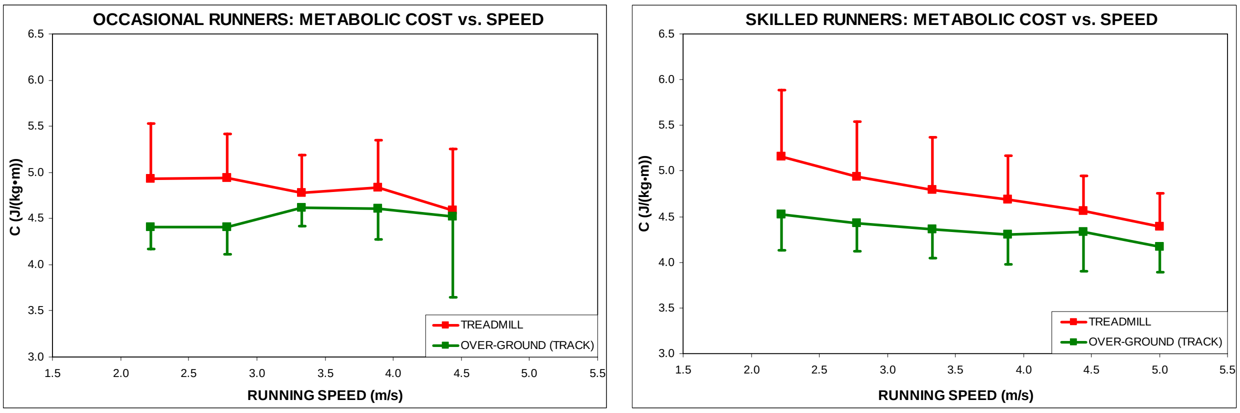

Figure 19.8. Metabolic cost as a function of running speed,
occasional (on the left) and skilled (on the right) runners, on treadmill and in the track.

In treadmill running, the statistical analysis has shown that there are no differences related to running levels, independently of speed (Pearce et al., 1983; Brisswalter et al., 1996; Maldonado et al., 2002; Dallam et al., 2005).

However, in the track running, C was found to be lower (albeit not significantly) in SR (4.39 ± 0.36 J/(kg·m)) than in OR (4.51 ± 0.39 J/(kg·m)), at all the investigated speeds. This is probably related to the subjects training levels: indeed, they could be not so different (Morgan et al., 1995).

Furthermore, comparing treadmill to track, independently of running level ($n = 63$) and speed, $R^2 = 0.6322$ and $r = 0.7951$. This means that no significant differences exist among these terrains.

5. DISCUSSION AND CONCLUSION

We come up to the conclusions that:

a) excepted metabolic cost, all other physiological variables increase linearly and similarly with running speed, independently of both running levels and types of terrain;

b) our cardio-respiratory values wholly concur with literature data (Margaria, 1938; Pearce et al., 1983; di Prampero, 1985; Williams, 1990; Kerdok, 2002; Riley, 2008);

c) as widely demonstrated in literature, metabolic cost is independent of speed and running level in each testing condition. The absence of a difference between treadmill and overground is probably due to similar biomechanical characteristics (i.e. stiffness) between the belt of the treadmill and the track;

d) independently of running abilities, metabolic cost is slightly greater on treadmill from 2.22 to 3.89 m/s ($p<0.05$);

e) finally, in the comparison among terrains, at the highest running speeds (e.g. ‘aerodynamic speed’), an important role could be played by acting forces (i.e. centripetal and centrifugal forces, air resistance).

The high correspondence between occasional and skilled runners is probably due to a non significant difference in training level: indeed, it seems that these two groups are related and comparable.

Therefore, in order to check the validity and applicability of these results, we suggest to:

- study the same variables in more trained subjects (for instance, top runners). Indeed, we expect that if a subject is much more trained, then he will be much more efficient and economical;
- when running occurs in the track, perform a more accurate check of the speed along curves;
- measure physiological parameters after a longer running time (i.e. after 1 hour);

- record the same variables during level walking to check whether a more variability occurs among various walking/running training levels;
- measure the important biomechanical variable of stiffness, as well;
- record and analyse both kinematics and kinetics in all the previous testing conditions.

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Chapter 20

BIOMECHANICS OF TREADMILL RUNNING

1. INTRODUCTION

1.1. Main biomechanical variables of treadmill running

Kinetics and muscle activity of treadmill locomotion (for explaining running economy) at different running speeds have already been investigated in many studies (Williams, 1985a; Lejeune et al., 1998; Kyrolainen et al., 2001; Kang et al., 2002; Seyfarth et al., 2002; Saunders et al., 2004; Dugan et al., 2005; Wheat et al., 2005). Instead, the main goal of this chapter is to focus on the kinematics of treadmill running (Collins et al., 2000).

To reach this aim, according to the rich literature, we investigated the behaviour of: a) some simple biomechanical variables (par. 2): stride frequency, stride length and duty factor; and b) some complex biomechanical variables (par. 3): mechanical external work, internal work and total work.

All these variables are fundamental both to characterize and to fully describe the mechanics of such a locomotion (see also chapter 1, par. 4). The mechanical ‘apparent’ efficiency is the last variable we examined (par. 4). It is derived from both a biomechanical variable (i.e. mechanical total work, W_{tot}) and a physiological measurement (i.e. metabolic cost, C).

In detail, biomechanical variables were measured discretely cycle by cycle, at the chosen sampling rate (100 Hz), in order to obtain average values (from each kinematic recording). Particularly, each variable has been elaborated by means of a custom-written LabVIEW software (Minetti et al., 1993; see also chapter 16, par. 2.2). Values of each anthropometric male parameter were taken from Winter’s work (2005: see also chapter 6, par. 2.1). Specifically, these kinematic (and biomechanical) variables were measured for all the subjects whose characteristics were already presented in chapter 16.

In detail, all simple and complex biomechanical variables (in all runners) are contained in the enclosed CD (Second Study, Chapter 20, Biomechanical variables and statistical analysis).

The last section focuses on the (absolute and relative) temporal and spatial (inter- and intra-) variability of the BCOM (par. 5) during treadmill running.

1.2. Statistical analysis

Statistical analysis was performed by using the average value of single biomechanical variable which was recorded during each running trial.

Results are presented as mean ± standard deviation (S.D.). The alpha test level set for statistical significance was 0.05.

The independent variable was progression speed (m/s), in each running group. The chosen dependent variables were the stride frequency (SF), the stride length (SL), the duty factor (DF), the external work (W_{ext}), the internal work (W_{int}) and the total work (W_{tot}).

A two-way ANOVA for related/unrelated measures (i.e. mixed design) was performed to compare the main effects of both speed and group. In addition, a post-*hoc* paired *t*-test with Bonferroni correction was used to detect the strength of the associations between single variables.

SPSS software (version 12.0 for Windows) was used for statistical analysis (Zakeri et al., 2006; Houdijk et al., 2009).

1.3. Graph legend

In each graph, the points represent mean values obtained by grouping the subjects performing the same running abilities (occasional runners *versus* skilled runners *versus* top runners) at different speeds.

Precisely, the average values of speed derived from the afore-mentioned *.vi *Motion Analysis Filter* in LabVIEW 2.2.1 (see also chapter 6, par. 2.1) have been considered (see Table 17.4, in chapter 17, par. 2). However, in order to avoid confusions, in the tables below the main speeds (2.22, 2.78, 3.33, 3.89, 4.44 and 5.00 m/s) are reported. Moreover, the average value of single kinematic recordings at each speed has been calculated.

The lines represent the simple graphic amalgamation of all the data and the vertical bars represent positive and negative standard deviations of the higher and lower speed curves (mean ± S.D.), respectively. Particularly:

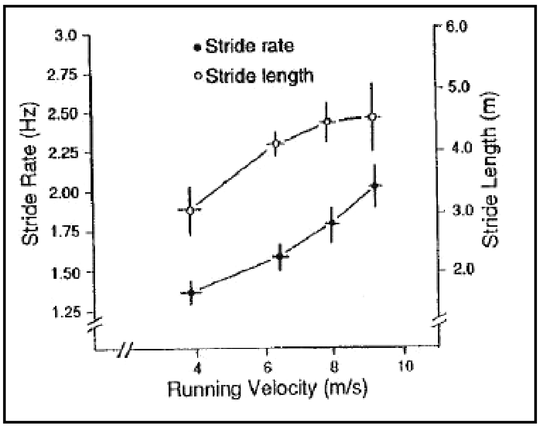
- red corresponds to occasional runners' performance;
- green corresponds to skilled runners' performance;
- ski-blue corresponds to top runners' performance.

2. SIMPLE BIOMECHANICAL VARIABLES

2.1. Stride frequency and stride length

Stride frequency and stride length were calculated as previously described in chapter 10 (par. 2 and 3, respectively).

As widely demonstrated in literature (Slocum et al., 1968; Cavanagh et al., 1977; Luhtanen et al., 1978; Cavanagh et al., 1982; Daniels, 1985; Williams, 1985a; Unnithan et al., 1990; Bailey et al., 1991; Brandon et al., 1992; Brisswalter et al., 1994; Morgan et al., 1994; Brisswalter et al., 1996; Derrick et al., 1998; De Wit et al., 2000; Kyrolainen et al., 2000; Sparrow, 2000; Bertram et al., 2001; Auvinet et al., 2002; Enoka, 2002; Mercer et al., 2002; Avogadro et al., 2003; Mercer et al., 2003; Saunders et al., 2004; Dallam et al., 2005; Dugan et al., 2005; McGinnis, 2005; Kilding et al., 2007; Steudel-Numbers et al., 2007; Chen et al., 2009; De Smet et al., 2009; Franz et al., 2009; Hanon et al., 2009), if stride length remains constant, then as stride frequency increases running speed increases; moreover, if stride frequency remains constant, speed increases as stride length increases (Figure 20.1. Average changes in stride length and stride frequency with running velocity, in Enoka (2002)).



For this reason, most runners naturally choose a stride length/stride frequency combination, which minimizes the metabolic cost (Laurent et al., 1986; Wirta et al., 1990; Sparrow, 2000; Danion et al., 2003; Kuo et al., 2005; Hunter et al., 2007; see also chapter 1, par. 2.4.3). This preferred stride length/stride frequency results in what might be termed *self-optimization*.

Furthermore, it is important to note that the contribution of changes in stride length and stride frequency to running velocity are different at low and high speeds (Williams, 1985a; Derrick et al., 1998; Mercer et al., 2003; De Smet et al., 2009). In general, at the higher speeds, runners increase much more stride length than stride frequency. This pattern is independent of running abilities: it is probably due to a less energy required to lengthen the stride within a reasonable limit than to increase stride frequency (Slocum et al., 1968; Cavanagh et al., 1982; Alexander, 1992; Derrick et al., 1998; Kyrolainen et al., 2000; Auvinet et al., 2002; Enoka, 2002; Zatsiorsky, 2002; Dallam et al., 2005; Dugan et al., 2005; Kirtley, 2006; De Smet et al., 2009; Leskinen et al., 2009).

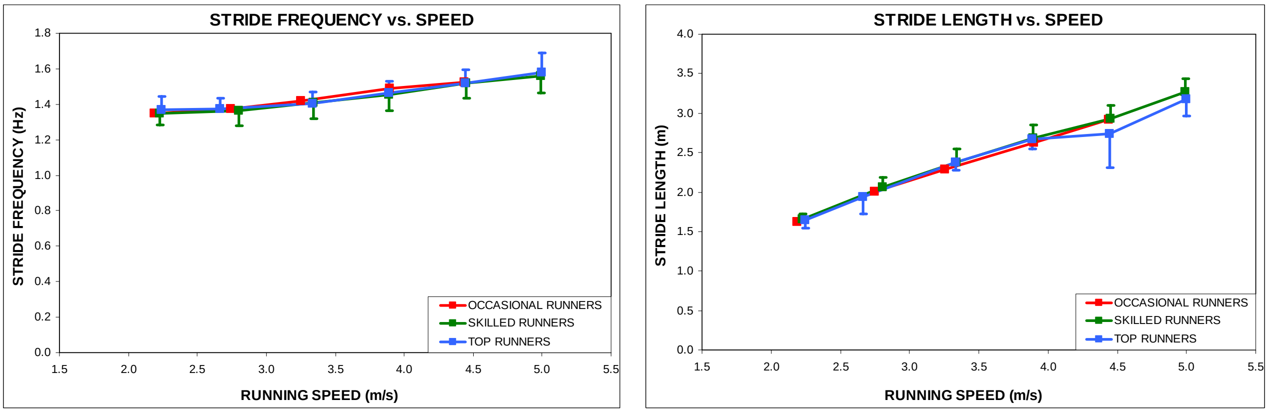


Figure 20.2. Stride frequency (on the left) and stride length (on the right) as a function of running speed.

As shown in Figure 20.2 and Table 20.1a, our results agree with this conclusion in all runners.

SPEED (m/s)	OCCASIONAL RUNNERS		SKILLED RUNNERS		TOP RUNNERS	
	SF (Hz)	SL (m)	SF (Hz)	SL (m)	SF (Hz)	SL (m)
2.22	1.35 ± 0.07	1.63 ± 0.06	1.35 ± 0.06	1.66 ± 0.07	1.37 ± 0.07	1.64 ± 0.09
2.78	1.37 ± 0.06	2.00 ± 0.05	1.36 ± 0.08	2.06 ± 0.13	1.38 ± 0.06	1.94 ± 0.22
3.33	1.42 ± 0.06	2.29 ± 0.07	1.41 ± 0.09	2.38 ± 0.17	1.40 ± 0.06	2.38 ± 0.10
3.89	1.47 ± 0.07	2.62 ± 0.11	1.45 ± 0.09	2.68 ± 0.17	1.46 ± 0.07	2.66 ± 0.12
4.44	1.51 ± 0.08	2.94 ± 0.19	1.52 ± 0.09	2.93 ± 0.16	1.52 ± 0.08	2.74 ± 0.42
5.00	result no available		1.56 ± 0.09	3.26 ± 0.17	1.58 ± 0.11	3.17 ± 0.21

Table 20.1a. SF and SL in occasional, skilled and top runners.

By the qualitative analysis of these graphs, our results show that:

- on average, stride frequency is similar in all groups (Hubbard, 1939; James et al., 1973; Cavanagh et al., 1977; Avogadro et al., 2003). In fact, it ranges from a minimum of 1.35 ± 0.07 Hz in OR, 1.35 ± 0.06 Hz in SR and 1.37 ± 0.07 Hz in TR to a maximum of 1.51 ± 0.08 Hz in OR (at 4.44 m/s), 1.56 ± 0.09 Hz in SR and 1.58 ± 0.11 Hz in TR (at 5.00 m/s). According to literature (Enoka, 2002; Mercer et al., 2002), in each group SF slightly increases across speeds ($p<0.05$; $n = 7$, $R^2 = 0.9922$ and $r = 0.9961$ in OR; $n = 7$, $R^2 = 0.9765$ and $r = 0.9800$ in SR; $n = 5$, $R^2 = 0.9513$ and $r = 0.9753$ in TR);
- on average, stride length is quite similar in all groups (Hubbard, 1939; James et al., 1973; Cavanagh et al., 1977; Derrick et al., 1998). In fact, it ranges from a minimum of 1.63 ± 0.06 m in OR, 1.66 ± 0.07 m in SR and 1.64 ± 0.09 m in TR to a maximum of 2.94 ± 0.19 m in OR (at 4.44 m/s), 3.26 ± 0.17 m in SR and 3.17 ± 0.21 m in TR (at 5.00 m/s) (Boje, 1944). According to literature (Cavanagh et al., 1990; Mercer et al., 2002), in each group SL increases across speeds ($p<0.05$; $R^2 = 0.9969$ and $r = 0.9984$ in OR; $R^2 = 0.9956$ and r 0.9978 in SR; $R^2 = 0.9753$ and $r = 0.9875$ in TR).

The statistical analysis supports this preliminary analysis, showing that there are no significant differences in each of these biomechanical variables between occasional, skilled and top runners (Table 20.1b).

STRIDE FREQUENCY (SF) STRIDE LENGTH (SL)	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
OCCASIONAL RUNNERS		p=NS	p=NS
SKILLED RUNNERS	p=NS		p=NS
TOP RUNNERS	p=NS	p=NS	

Table 20.1b. Statistical comparison in SF and SL among runner groups.

Importantly, these differences are marked as running speeds increase ($p<0.01$) (Table 20.1c).

SUBJECTS	STRIDE FREQUENCY and STRIDE LENGTH
OCCASIONAL RUNNERS	$p<0.05$ comparing each speed
SKILLED RUNNERS	$p<0.05$ comparing each speed
TOP RUNNERS	$p<0.05$ comparing each speed

Table 20.1c. SF and SL as a function of running speed.

2.2. Duty factor

Duty factor was calculated as previously described in chapter 6 (par. 2.1) and 10 (par. 5).

All duty factor values are graphically represented in Figure 20.3 and completely reported in Table 20.2a.

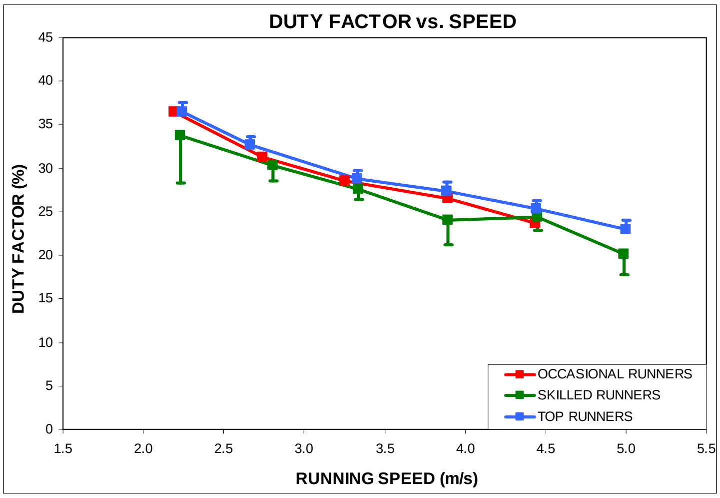


Figure 20.3. Duty factor as a function of running speed.

SPEED (m/s)	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
	DF (%)	DF (%)	DF (%)
2.22	36.47 ± 3.95	33.70 ± 5.34	36.51 ± 1.78
2.78	31.22 ± 2.74	30.26 ± 1.68	32.63 ± 2.89
3.33	28.56 ± 2.39	27.60 ± 1.24	28.72 ± 2.49
3.89	26.58 ± 2.65	24.07 ± 2.86	27.40 ± 1.38
4.44	23.65 ± 2.05	24.35 ± 1.51	25.32 ± 0.90
5.00	result no available	20.13 ± 2.36	23.01 ± 2.15

Table 20.2a. DF in occasional, skilled and top runners.

As shown in Figure 20.3, duty factor is reduced with increased speed ($p<0.001$), according to literature data (Slocum et al., 1968; Alexander, 1992; Minetti, 1998; Saibene et al., 2003; Hoyt et al., 2006). This pattern is not dependent on running abilities (Table 20.2b).

SUBJECTS	DUTY FACTOR (DF)
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed
TOP RUNNERS	$p<0.001$ comparing each speed

Table 20.2b. DF as a function of running speed.

The qualitative analysis shows that duty factor is slightly higher in OR (29.29 ± 2.75%, independently of speed) and TR (29.93 ± 1.93%) than in SR (25.51 ± 2.02%).

Yet, the statistical analysis does not support these results, showing that there are no significant differences among different running abilities (Table 20.2c).

DUTY FACTOR (DF)	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
OCCASIONAL RUNNERS		p=NS	p=NS
SKILLED RUNNERS	p=NS		p=NS
TOP RUNNERS	p=NS	p=NS	

Table 20.2c. Statistical comparison in DF among runner groups.

3. COMPLEX BIOMECHANICAL VARIABLES

3.1. Mechanical external work and mechanical internal work

Mechanical external work (W_{ext}) and mechanical internal work (W_{int}) were calculated as previously described in chapter 6 (par. 2.1) and 10 (par. 6 and 8).

All external and internal work values are graphically represented in Figure 20.4 and completely reported in Table 20.3a.

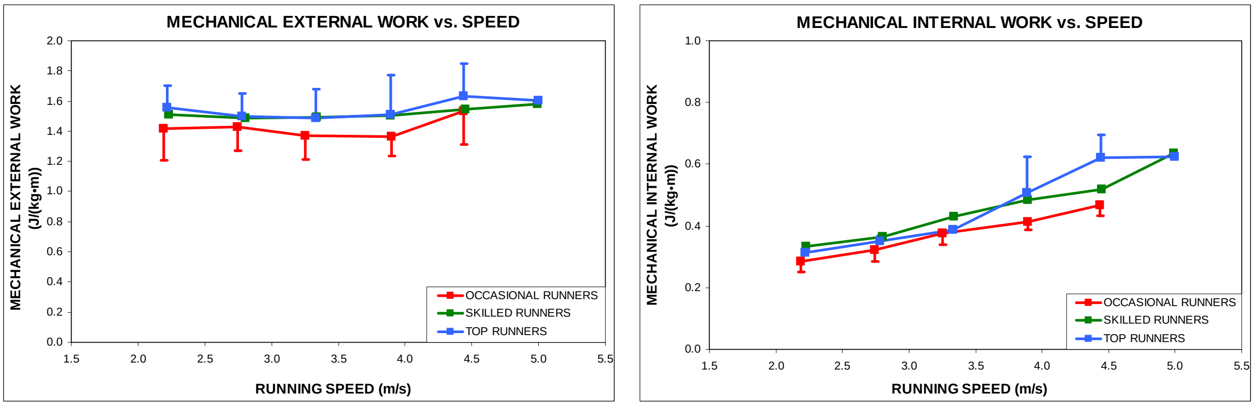


Figure 20.4. Mechanical external work (on the left) and mechanical internal work (on the right) as a function of running speed.

SPEED (m/s)	**OCCASIONAL RUNNERS**		**SKILLED RUNNERS**		**TOP RUNNERS**	
	W_{ext} (J/(kg·m))	**W_{int} (J/(kg·m))**	**W_{ext} (J/(kg·m))**	**W_{int} (J/(kg·m))**	**W_{ext} (J/(kg·m))**	**W_{int} (J/(kg·m))**
2.22	1.42 ± 0.21	0.28 ± 0.03	1.51 ± 0.15	0.33 ± 0.05	1.49 ± 0.11	0.30 ± 0.01
2.78	1.43 ± 0.15	0.32 ± 0.04	1.49 ± 0.09	0.36 ± 0.04	1.46 ± 0.09	0.33 ± 0.02
3.33	1.37 ± 0.16	0.38 ± 0.04	1.49 ± 0.13	0.43 ± 0.09	1.43 ± 0.11	0.39 ± 0.03
3.89	1.36 ± 0.12	0.41 ± 0.02	1.50 ± 0.17	0.48 ± 0.10	1.43 ± 0.15	0.45 ± 0.09
4.44	1.54 ± 0.22	0.47 ± 0.03	1.54 ± 0.20	0.52 ± 0.16	1.46 ± 0.13	0.51 ± 0.08
5.00	result no available		1.56 ± 0.09	0.63 ± 0.17	1.54 ± 0.14	0.56 ± 0.07

Table 20.3a. W_{ext} and W_{int} in occasional, skilled and top runners.

The qualitative analysis shows that W_{ext} (left graph) is slightly higher in SR (1.52 ± 0.15 J/(kg·m), independently of speed) and TR (1.47 ± 0.12 J/(kg·m)) than in OR (1.42 ± 0.17 J/(kg·m)). At each running speed, between skilled and top runners, no evident differences are found.

Moreover, W_{int} (right graph) is slightly greater in SR (0.46 ± 0.10 J/(kg·m), independently of speed) and TR (0.42 ± 0.05 J/(kg·m)) than in OR (0.37 ± 0.07 J/(kg·m)). This pattern occurs more evidently at the highest speeds. At each running speed, between skilled and top runners, no evident differences are found, as well.

In external work, the statistical comparison among the three groups has shown that, at all the investigated speeds, occasional runners slightly differ to both skilled and top runners (Table 20.3b).

EXTERNAL WORK (W_{ext})	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
OCCASIONAL RUNNERS		$p<0.05$	$p<0.05$
SKILLED RUNNERS	$p<0.05$		
TOP RUNNERS	$p<0.05$		

Table 20.3b. Statistical comparison in W_{ext} among runner groups.

However, in internal work, occasional runners slightly differ to the other two groups only at the highest running speeds (Table 20.3c).

INTERNAL WORK (W_{int})	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
OCCASIONAL RUNNERS		$p<0.05$	$p<0.05$
SKILLED RUNNERS	$p<0.05$ (from 3.89 to 4.44 m/s)		
TOP RUNNERS	$p<0.05$ (from 3.89 to 4.44 m/s)		

Table 20.3c. Statistical comparison in W_{int} among runner groups.

Furthermore, in each group (independently of running abilities), external work does not significantly change with speed (Table 20.3d; according to Avogadro et al., 2004).

SUBJECTS	**EXTERNAL WORK (W_{ext})**
OCCASIONAL RUNNERS	p=NS comparing each speed
SKILLED RUNNERS	p=NS comparing each speed
TOP RUNNERS	p=NS comparing each speed

Table 20.3d. W_{ext} as a function of running speed.

However, independently of running abilities, internal work slightly increases with speed ($p<0.05$, according to Avogadro et al., 2003) (Table 20.3e).

SUBJECTS	**INTERNAL WORK (W_{int})**
OCCASIONAL RUNNERS	$p<0.05$ comparing each speed
SKILLED RUNNERS	$p<0.05$ comparing each speed
TOP RUNNERS	$p<0.05$ comparing each speed

Table 20.3e. W_{int} as a function of running speed.

Our results concur with literature data (Cavagna et al., 1964; 1976; 1983; Cavanagh et al., 1985a; Cavagna et al., 1986; Willems et al., 1995; Minetti, 1998; Avogadro et al., 2003; Saibene et al., 2003; Leskinen et al., 2009).

3.2. Fourier analysis to calculate mechanical external work

The mathematical approach widely described in chapter 11 permits us:

- to demonstrate again that kinetic (KE) and potential (PE) energy increases are in phase, according to literature data (Figure 20.5, graph above. For more details about the legend graph see also chapter 11, par. 1.3);
- finally, to calculate external work also using Fourier coefficients (amplitudes and phases; Figure 20.5, graph below). All external work values are completely reported in Table 20.4a.

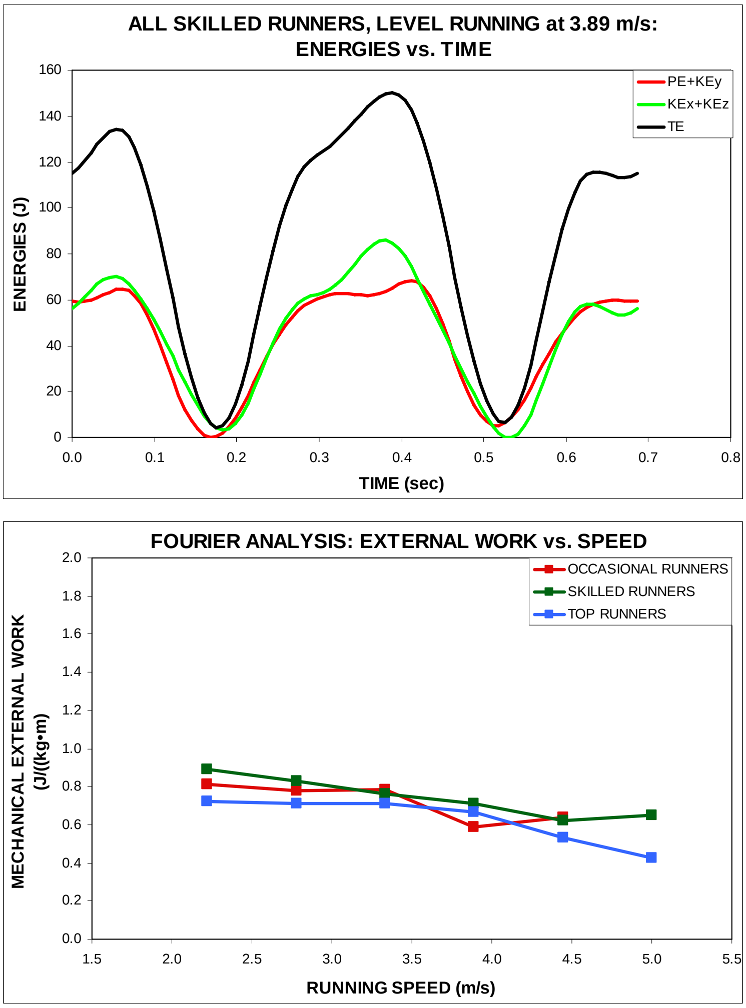


Figure 20.5. Energies *versus* time (graph above) and external work derived by Fourier analysis as a function of running speed (graph below).

FOURIER ANALYSIS	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
SPEED (m/s)	**W_{ext} (J/(kg·m))**	**W_{ext} (J/(kg·m))**	**W_{ext} (J/(kg·m))**
2.22	0.814	0.892	0.768
2.78	0.776	0.830	0.744
3.33	0.784	0.763	0.717
3.89	0.588	0.713	0.665
4.44	0.641	0.621	0.595
5.00	result no available	0.652	0.480

Table 20.4a. W_{ext} derived by Fourier analysis in occasional, skilled and top runners.

The external work values derived by Fourier analysis are very smaller than corresponding values derived by kinematic analysis (cycle by cycle), independently of running ability and speed ($p<0.001$). Moreover, TR have the smallest values (0.662 J/(kg·m), independently of speed) than OR (0.721 J/(kg·m)) and SR (0.745 J/(kg·m)).

This result has been confirmed by the statistical analysis (Table 20.4b), as well.

W_{ext} (from Fourier analysis)	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
OCCASIONAL RUNNERS			$p<0.001$
SKILLED RUNNERS			$p<0.001$
TOP RUNNERS	$p<0.01$	$p<0.01$	

Table 20.4b. Statistical comparison in W_{ext} derived by Fourier analysis among runner groups.

On average, as a function of speed, W_{ext} decreases linearly with running speed, independently of subject ability. However, both in occasional and top runners, this pattern is more evident at the highest speeds (Table 20.4c). As previously demonstrated (par. 3.1) our result concurs with literature data.

SUBJECTS	**FOURIER ANALYSIS EXTERNAL WORK (W_{ext})**
OCCASIONAL RUNNERS	$p<0.001$ from 3.33 to 3.89 m/s
SKILLED RUNNERS	$p<0.001$ comparing each speed
TOP RUNNERS	$p<0.001$ from 3.33 to 5.00 m/s

Table 20.4c. W_{ext} derived by Fourier analysis as a function of running speed.

In detail, single mechanical external work and % recovery values (in all runners) are contained in the enclosed CD (Second Study, Chapter 20, Spreadsheet W_{ext} and %Recovery).

3.3. Mechanical total work

Mechanical total work (W_{tot}) was calculated as the sum of W_{ext} and W_{int}, which are considered as two separate entities as previously described in chapter 6 (par. 2.1) and 10 (par. 9).

All total work values are graphically represented in Figure 20.6 and completely reported in Table 20.5a.

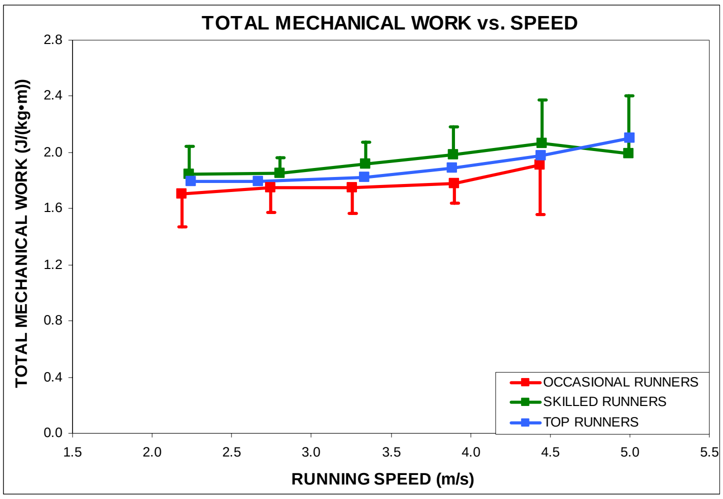


Figure 20.6. Mechanical total work as a function of running speed.

SPEED (m/s)	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
	W_{tot} (J/(kg·m))	W_{tot} (J/(kg·m))	W_{tot} (J/(kg·m))
2.22	1.70 ± 0.23	1.84 ± 0.20	1.79 ± 0.12
2.78	1.75 ± 0.18	1.85 ± 0.11	1.80 ± 0.10
3.33	1.75 ± 0.18	1.86 ± 0.15	1.82 ± 0.11
3.89	1.78 ± 0.13	1.97 ± 0.19	1.89 ± 0.23
4.44	1.91 ± 0.52	2.06 ± 0.31	1.98 ± 0.19
5.00	result no available	1.99 ± 0.41	2.10 ± 0.19

Table 20.5a. W_{tot} in occasional, skilled and top runners.

Both qualitative and statistical analysis show that W_{tot} is slightly higher ($p<0.05$) in SR (1.94 ± 0.23 J/(kg·m), independently of speed) and TR (1.89 ± 0.16 J/(kg·m)) than in OR (1.77 ± 0.22 J/(kg·m)), as already stated in W_{ext} and W_{int} (Table 20.5b).

TOTAL WORK (W_{tot})	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
OCCASIONAL RUNNERS		$p<0.05$	$p<0.05$
SKILLED RUNNERS	$p<0.05$		
TOP RUNNERS	$p<0.05$		

Table 20.5b. Statistical comparison in W_{tot} among runner groups.

The mechanical total work does not significantly change with speed (Burdett et al., 1983; Cavagna et al., 1983; Casaburi et al., 1989; Willems et al., 1995; Bianchi et al., 1998; Candau et al., 1998; Avogadro et al., 2003; Borrani et al., 2003), independently of running abilities (Table 20.5c).

SUBJECTS	TOTAL WORK (W_{tot})
OCCASIONAL RUNNERS	p=NS comparing each speed
SKILLED RUNNERS	p=NS comparing each speed
TOP RUNNERS	p=NS comparing each speed

Table 20.5c. W_{tot} as a function of running speed.

4. MECHANICAL 'APPARENT' EFFICIENCY

4.1. Definition and main characteristics

Mechanical 'apparent' efficiency (Minetti, 2004) η is a dimensionless variable widely analysed in literature (Pugh, 1971; Lloyd et al., 1972; Asmussen et al., 1974; Shepard, 1975; Aruin et al., 1979; Ito et al., 1983; Cavanagh et al., 1985a; Sparrow et al., 1987; Kaneko, 1990; Caldwell et al., 1992; Sun et al., 1993; Kyrolainen et al., 1995; Ettema, 1996; Bosco et al., 1997; Woledge, 1997; Sparrow, 2000; Zamparo et al., 2000; Minetti et al., 2001; Preedy et al., 2001; Terrier et al., 2001; Frost et al., 2002; Detrembleur et al., 2003; Biewener et al., 2004; Robertson et al., 2004; Saunders et al., 2004; Ardigò et al., 2005; Stoquart et al., 2005; Winter, 2005; Mian et al., 2006; van de Hecke et al., 2007; Umberger et al., 2007; Bar-Haim et al., 2008; Capelli et al., 2008; Sasaki et al., 2008; Zamparo et al., 2008; Mahaudens et al., 2009; Nakai et al., 2009; Steudel-Numbers et al., 2009). To avoid terminology confusion, it has to be distinguished to muscular efficiency, muscle efficiency, mechanical or work efficiency (Shepard, 1975; Cavanagh et al., 1985b; Williams, 1985b; Sparrow, 2000; Winter, 2005).

In studies of human locomotion (i.e. a biological system; Sparrow et al., 1987; Kaneko, 1990; Sun et al., 1993; Ettema, 1996; Pereira et al., 1997; Sparrow, 2000; Frost et al., 2002; Robertson et al., 2004; Capelli et al., 2008; Nakai et al., 2009), mechanical 'apparent' efficiency has commonly been defined as:

$$\eta = \frac{W_{tot}}{C} \quad \text{[Eq. 20.1]}$$

where W_{tot} constitutes the mechanical total work performed during running (sum of W_{ext} and W_{int}) and C is the corresponding metabolic cost. For more information about these two variables, see also chapter 10, par. 9 (Lloyd et al., 1972) and 17, par. 1, respectively.

Numerous factors (Kyrolainen et al., 2000) influence the mechanical 'apparent' efficiency of running, such as: a) age (Daniels et al., 1978); b) gender (Bransford et al., 1977); c) air resistance (Costill et al., 1969); d) body mass (Cureton et al., 1978); e) maximal aerobic power; f) muscle fibre distribution (Bosco et al., 1987); and g) resting metabolic rate (Winter, 2005). Finally, variations in efficiency, can be attributed to methodological differences (Kaneko, 1990).

4.2. Mechanical 'apparent' efficiency in our results

Therefore, the knowledge of both mechanical total work and metabolic cost permits us to calculate the mechanical 'apparent' efficiency in each runner group.

All mechanical 'apparent' efficiency values are graphically represented in Figure 20.7 and completely reported in Table 20.6a.

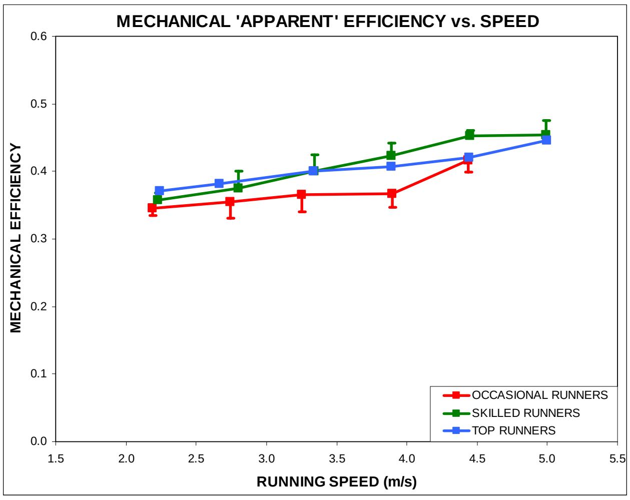


Figure 20.7. Mechanical 'apparent' efficiency as a function of running speed.

SPEED (m/s)	OCCASIONAL RUNNERS	SKILLED RUNNERS	TOP RUNNERS
	η	η	η
2.22	0.386 ± 0.010	0.407 ± 0.011	0.371 ± 0.008
2.78	0.397 ± 0.023	0.418 ± 0.025	0.382 ± 0.013
3.33	0.378 ± 0.026	0.440 ± 0.024	0.400 ± 0.007
3.89	0.386 ± 0.021	0.461 ± 0.018	0.408 ± 0.005
4.44	0.422 ± 0.017	0.476 ± 0.019	0.421 ± 0.011
5.00	result no available	0.454 ± 0.022	0.446 ± 0.017

Table 20.6a. η in occasional, skilled and top runners.

As shown in Figure 20.7, the qualitative analysis has shown that SR (0.410 ± 0.018, independently of speed) are slightly more efficient than OR (0.393 ± 0.019) and TR (0.404 ± 0.001). This result seems to confirm literature data (Kaneko, 1990; Capelli et al., 2008).

However, the statistical analysis (differences both among speed and group were assessed by using a one-way ANOVA for unrelated measures with a post-*hoc* paired *t*-test, with Bonferroni correction) shows that:

- at the lowest speeds (from 2.22 to 2.78 m/s), there are no significant differences among the groups (Table 20.6b);
- both skilled runners and top runners become more efficient as running speed increases ($p<0.001$ and $p<0.01$, respectively; Kaneko, 1990) (Table 20.6c);
- on average, skilled runners are more efficient than top runners ($p<0.01$). The only exception is the highest speed of 5.00 m/s.

MECHANICAL 'APPARENT' EFFICIENCY (η)	**OCCASIONAL RUNNERS**	**SKILLED RUNNERS**	**TOP RUNNERS**
OCCASIONAL RUNNERS		$p<0.001$	$p<0.01$
SKILLED RUNNERS	$p<0.001$ (from 2.78 to 4.44 m/s)		
TOP RUNNERS	$p<0.01$ (from 2.78 to 4.44 m/s)		

Table 20.6b. Statistical comparison in η among runner groups.

SUBJECTS	**MECHANICAL 'APPARENT' EFFICIENCY (η)**
OCCASIONAL RUNNERS	$p<0.001$ comparing each speed
SKILLED RUNNERS	$p<0.001$ comparing each speed
TOP RUNNERS	$p<0.001$ comparing each speed

Table 20.6c. η as a function of running speed.

In general, a possible explanation of such limited differences could be found in Kyrolainen et al. (1995): 'the subject groups did not differ so significantly in mechanical efficiency due to high inter-individual variance among subject groups'.

In detail, single mechanical apparent efficiency (in all runners) are contained in the enclosed CD (Second Study, Chapter 20, Mechanical apparent efficiency).

5. SPATIAL-TEMPORAL VARIABILITY OF THE BODY CENTRE OF MASS

5.1. Introduction

The importance of movement variability (i.e. for sports biomechanics) was overviewed by some researchers (Skaggs et al., 2000; Bartlett et al., 2007). Their studies investigated both inter- and intra- individual movement variability in some sports (i.e. javelin and discus throwing, basketball shooting and locomotion).

But '*could this movement variability occur also in the displacement of body centre of mass?*' Therefore, '*what are the main characteristics of such a variability?*'

Focusing on studying the mechanical variability of the BCOM in human locomotion (i.e. running), it has been demonstrated that the most commonly used method is film (or video) analysis (Cavanagh et al., 1977; Bates et al., 1979; Belli et al., 1995).

The importance of recording the variability of BCOM has been emphasized in order:

a) to better understand the biomechanics of human locomotion, as stated in Bartlett et al. (2007) and Beauchet et al. (2009b);
b) to verify whether significant differences among runners featuring different abilities exist;
c) to state whether a relationship exists between biomechanical factors (i.e. spatial/temporal variability of the BCOM) and metabolic findings (i.e. running economy).

As a result, the present section has been designed to study intra- and inter- individual mechanical step variability (see par. 5.2 onwards) of runners measuring both:

1. the delta time period (Δt), related to the temporal variability of the BCOM;
2. the delta BCOM displacement (ΔBCOM), related to the spatial variability of the BCOM.

5.2. Mechanical step variability

In the recent past, the intra-individual step mechanical variability has been studied by measuring both vertical displacement of the body and step time parameters by means of a kinematic arm in runners experienced on long-distance running or in sprint running (Belli et al., 1993; 1995). In detail, absolute and relative means and standard deviations of 70-120 steps (based upon stride frequency) were measured. It has been found that: a) absolute step time and vertical displacement of the BCOM decreased linearly with the increase of velocity; b) absolute standard deviation of these variables does not change as a function of running speed; c) absolute step time increased with velocity level; d) however, absolute vertical displacement increased only in some velocity ranges; e) relative step time and vertical displacement variability increased clearly with velocity level; f) the asymmetry between two successive legs on maximal flexion was highly negatively related to the corresponding differences in vertical propulsion; finally, g) at sub-maximal levels, a significant relationship was found between average cost of running and mean of the vertical displacement of the BCOM: the higher the vertical displacement variability, the higher the energy cost of running.

More recently, especially in walking gait, by investigating the relationship between stride time variability and speed, it has been demonstrated that: a) gait variability increases while walking speed decreases (Beauchet et al., 2009b); b) thus, a decreases in walking speed could be a potential confounder in evaluating gait variability (Beauchet et al., 2009b); and c) low and high spatio-temporal stride-to-stride variability may reflect gait stability in healthy people (Beauchet et al., 2009a).

Differently to these pioneer studies, we measured and recorded mechanical step variability (both spatial and temporal) of the BCOM (not only of isolated segments such as the head, the hip or the free hand) along all movement directions (forward, vertical and lateral) by the motion capture system. Therefore, the following sections will focus on the specific analysis we applied and the corresponding results we obtained.

5.3. Mechanical step variability of the body centre of mass in our study

Temporal and spatial variability of the BCOM have been measured only in occasional runners and skilled runners involved in the second study (see also chapter 16). Top runners were not considered in this analysis because the poor number of subjects we investigated.

Specifically, mechanical step variability has been analysed as a function of both:

1. <u>speed</u>, independently of running level;
2. <u>subject</u>, as running speed changes.

In detail, the 3D displacement of the BCOM has been obtained by means of the aforementioned custom-written LabVIEW software (see also chapter 6, par. 2.1), and its 3D trajectory has been then analysed by means of the software Acqknowledge® (version 3.9, XP, UK; Mikhov et al., 1998; Lescot et al., 2005; Acqknowledge Software Guide, 2008).

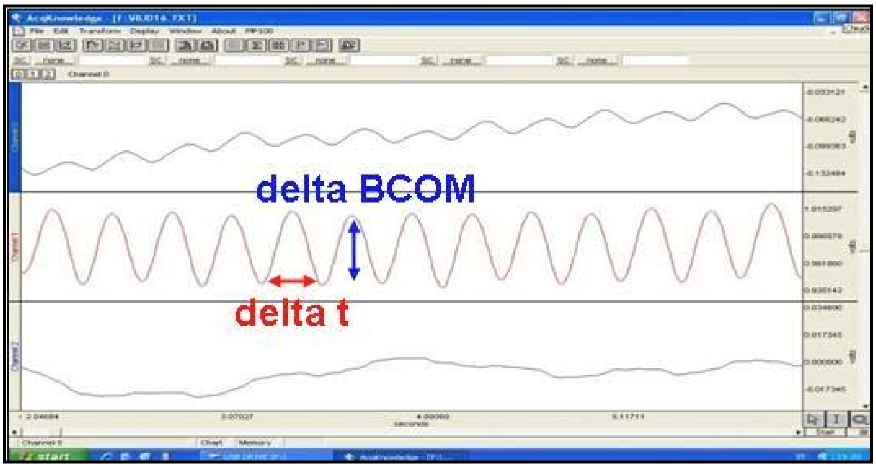


Figure 20.8. Three-dimensional displacement of the BCOM, visualised by the software Acqknowledge.

Acqknowledge software has been used to visualize and quantify, in each movement direction (channel 0 corresponds to forward direction; channel 1 to vertical; and channel 2 to lateral), the main parameters of interest: a) the maximal displacement of the BCOM (MAX, in m); and b) the minimal displacement of the BCOM (min, in m). Furthermore, the difference between the maximal (MAX) and the minimal (min) displacement value in a step defined the **spatial variability** (ΔBCOM). It was calculated in each movement direction. The **temporal variability** in vertical direction (corresponding to the duration of each step, Δt, in s) was also calculated (Figure 20.8), according to Belli et al. (1995). The variability of such parameters (Δt and ΔBCOM) was assessed over 10 consecutive running steps for each of the three kinematic registrations (see also chapter 16, par. 2). In each recordings, average values have been then obtained.

In detail, single files from Acqknowledge (in all runners) are contained in the enclosed CD (Second Study, Chapter 20, Software Acqknowledge, Template Acqknowledge and variability of the BCOM).

5.4. Data analysis of mechanical step variability

Means (M), standard deviations (S.D.) and coefficients of variation (CV) of each mechanical variable were computed in each running trial (Belli et al., 1995; Danion et al., 2003). Only speeds from 2.22 to 4.44 m/s were considered in our analysis because the poor number of subjects who were able to complete all the test protocol (see also chapter 16). To be precise:

a) **absolute variability** was assessed by calculating means (M). On whole, 37 trials have to be rejected because of some reasons: for instance, an insufficient number of stride cycles, the presence of a noise in the kinematic signal and so on. For more information, see also chapter 16 (par. 2) and 17 (par. 2.1);
b) **relative variability** was assessed by calculating the coefficient of variation (CV). It seems to be important because, in each duration, it was regarded as an index of stability of gait within subjects (Maruyama et al., 1992). On whole, 72 trials have to be rejected because some reasons (see also chapter 16, par. 2, and 17, par. 2.1).

Both absolute and relative variability of the BCOM have been calculated by considering means or coefficients of variation *per* speed (inter-individual variability) and *per* subject.

The pattern of both spatial (3D displacement) and temporal (step duration) variability of the BCOM has been analysed as a function of running speed and metabolic cost (Belli et al., 1995). Moreover, they have been analysed as a function of mechanical ‘apparent’ efficiency (see par. 4 above), as well (Belli et al., 1995).

5.5. Statistical analysis

We were interested in examining the value of R^2 (Determination Coefficient and corresponding value of r, Correlation Coefficient) in order to verify if linear regression could be a reasonable function representing each testing condition. To state if a specific value of r is satisfactory, we used the proper *Scientific Tables of Correlation Coefficient r to verify r = 0*. Consequently, knowing both the *r* value and the total number of values (*n - 1*), it becomes possible to define the significance of each specific regression analysis. The alpha test level for statistical significance was 0.05.

Therefore, linear regressions were used to state the significance of relationship between Δt or ΔBCOM and metabolic cost or mechanical efficiency. According to trials rejected:

a) in both absolute and relative variability, independently of subjects, n = 5;
b) in absolute variability, independently of subjects: in forward direction, n = 63 (ΔBCOM) and n = 65 (Δt); in vertical direction, n = 65 (ΔBCOM and Δt); in lateral direction, n = 59 (ΔBCOM) and n = 65 (Δt);
c) in relative variability, independently of subjects: in forward direction, n = 56 (ΔBCOM) and n = 61 (Δt); in vertical direction, n = 54 (ΔBCOM) and n = 57 (Δt); in lateral direction, n = 57 (ΔBCOM and Δt);
d) in both absolute and relative variability, independently of speed, n = 14 because subjects are studied independently of their running ability.

5.6. Results of our study

Being able to perform a detailed kinematic analysis and simultaneously measure the metabolic requirement, makes it possible to study the relationship between the temporal and spatial variability of the BCOM, the metabolic cost and the mechanical 'apparent' efficiency of running.

As a result, the cumulative analysis of these parameters could help in understanding whether the differences in C and η among individuals with different running abilities can be attributed, among the others, to the variability of biomechanical factors (i.e. mechanical step variability of the BCOM).

Therefore, in the following sections we will focus on both absolute and relative variability of the BCOM when data (means and coefficients of variation) has been analysed *per* speed or *per* subject.

5.6.1. Absolute variability of the body centre of mass *per* speed

A. Means (M) of both temporal (Δt) and spatial (ΔBCOM) variability have been calculated as a function of speed, independently of running abilities. All values are reported in Table 20.7a:

SPEED (m/s)	M Δt	M ΔBCOM Forward direction	M ΔBCOM Vertical direction	M ΔBCOM Lateral direction
2.22	0.379 ± 0.024	0.014 ± 0.005	0.085 ± 0.026	0.015 ± 0.006
2.78	0.372 ± 0.029	0.016 ± 0.004	0.088 ± 0.026	0.015 ± 0.005
3.33	0.358 ± 0.023	0.016 ± 0.005	0.085 ± 0.023	0.014 ± 0.004
3.89	0.350 ± 0.027	0.015 ± 0.004	0.091 ± 0.018	0.014 ± 0.004
4.44	0.339 ± 0.023	0.016 ± 0.005	0.087 ± 0.016	0.014 ± 0.009

Table 20.7a. Means of variability of the BCOM *per* speed, independently of running abilities.

Precisely, our means show that:

- as illustrated in Figure 20.9 (left graph), step duration (Δt) highly decreases ($p<0.001$) with running speed ($n = 5$, $R^2 = 0.9950$, $r = 0.9975$) because the stride frequency increases with speed (see par. 2.1 above) simultaneously with the movement frequency of the BCOM. Importantly, this result concurs with Belli's data (1995);
- in spatial variability, no significant differences are found in both forward and vertical directions ($n = 5$, $R^2 = 0.4321$, $r = 0.6573$ and $n = 5$, $R^2 = 0.2302$, $r = 0.4798$, respectively). Negatively, this result does not concur with Belli's data (1995);
- however, as illustrated in Figure 20.9 (right graph), ΔBCOM in lateral direction highly decreases ($p<0.001$) with running speed ($n = 5$, $R^2 = 0.9757$, $r = 0.9878$).

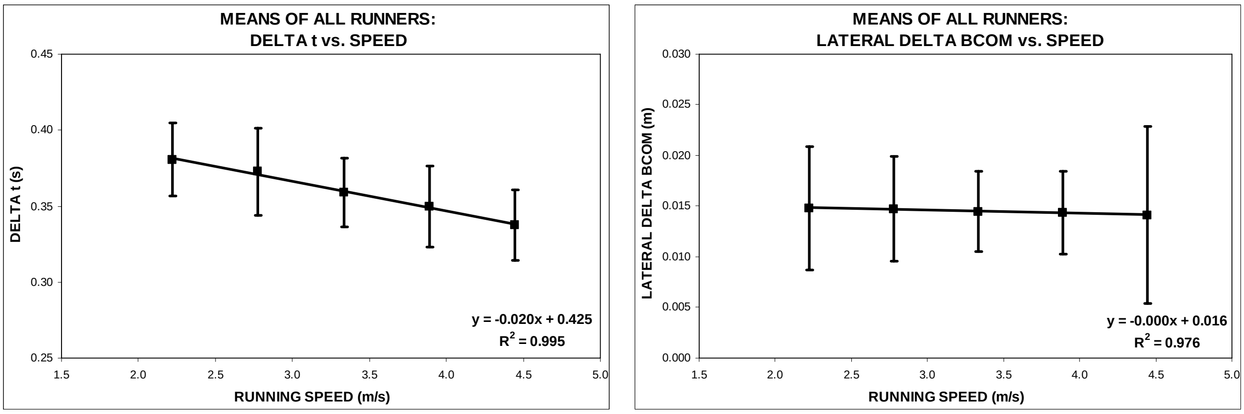


Figure 20.9. Delta t (on the left) and lateral delta BCOM (on the right) means as a function of running speed.

B. Average values of metabolic cost (± S.D.), independently of running ability, are presented in Table 20.7b:

SPEED (m/s)	**METABOLIC COST (J/(kg·m))**
2.22	4.966 ± 0.619
2.78	4.860 ± 0.484
3.33	4.736 ± 0.442
3.89	4.771 ± 0.530
4.44	4.721 ± 0.597

Table 20.7b. Average values of metabolic cost, independently of running abilities.

If the metabolic cost is considered as a function of the variability of the BCOM, our means show that:

- single values of both step duration (Δt) and metabolic cost (C) in all runners are graphically represented in Figure 20.10 (left graph). Precisely, metabolic cost slightly increases ($p<0.05$) with step duration of the BCOM ($n = 5$, $R^2 = 0.8029$, $r = 0.8960$) (right graph);

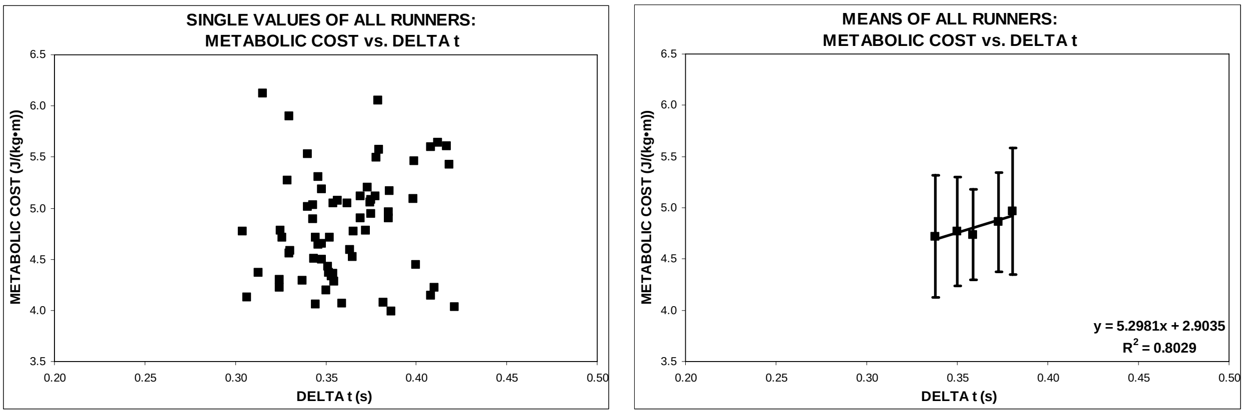


Figure 20.10. Single values of both delta t and metabolic cost in all runners (on the left) and metabolic cost means as a function of delta t means (on the right).

- single values of both forward variability (forward ΔBCOM) and metabolic cost (C) in all runners are graphically represented in Figure 20.11 (left graph). Precisely, metabolic cost slightly decreases ($p<0.05$) with forward displacement of the BCOM ($n = 5$, $R^2 = 0.7136$, $r = 0.8445$) (right graph);

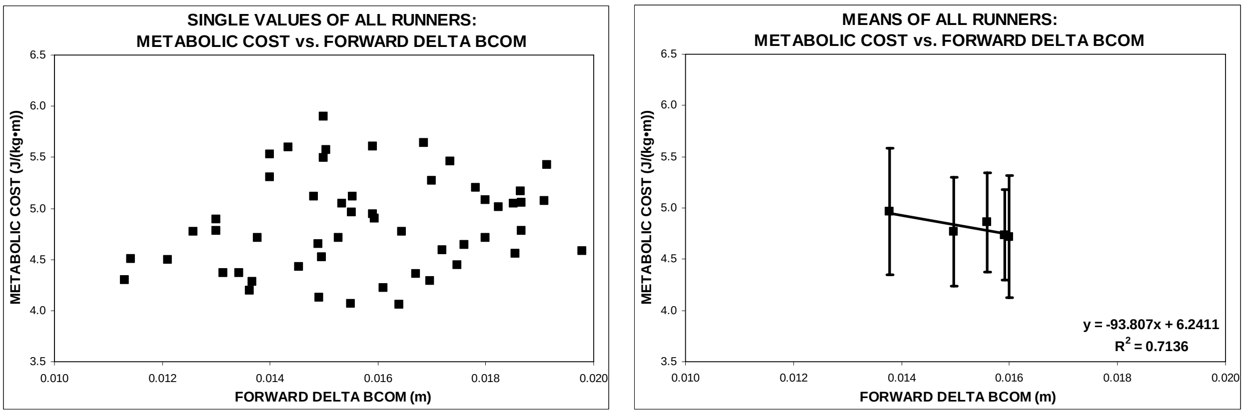


Figure 20.11. Single values of both forward delta BCOM and metabolic cost in all runners (on the left) and metabolic cost means as a function of forward delta BCOM means (on the right).

- differently, no significant differences are found in the vertical direction ($n = 5$, $R^2 = 0.0820$, $r = 0.2863$);
- finally, single values of both lateral variability (lateral ΔBCOM) and metabolic cost (C) in all runners are graphically represented in Figure 20.12 (left graph). Precisely, metabolic cost slightly increases ($p<0.05$) with lateral displacement of the BCOM ($n = 5$, $R^2 = 0.7820$, $r = 0.8843$) (right graph).

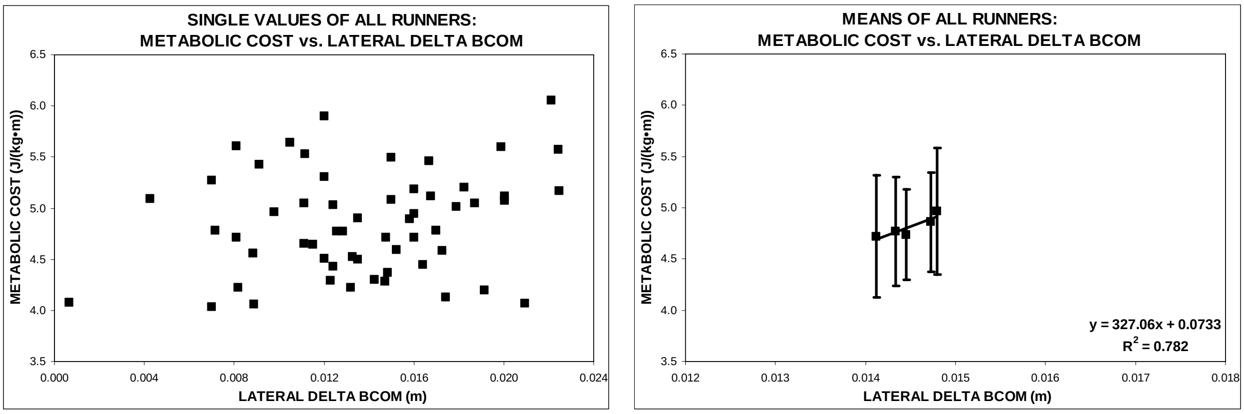


Figure 20.12. Single values of both lateral delta BCOM and metabolic cost in all runners (on the left) and metabolic cost means as a function of lateral delta BCOM means (on the right).

Negatively, all these results do not concur with Belli et al. (1995) data.

C. Average values of mechanical ‘apparent’ efficiency (± S.D.), independently of running ability, are presented in Table 20.7c:

SPEED (m/s)	MECHANICAL EFFICIENCY
2.22	0.397 ± 0.010
2.78	0.408 ± 0.024
3.33	0.410 ± 0.025
3.89	0.424 ± 0.019
4.44	0.449 ± 0.018

Table 20.7c. Average values of mechanical efficiency, independently of running abilities.

Precisely, our means show that:

- single values of both mechanical ‘apparent’ efficiency (η) and step duration (Δt) in all runners are graphically represented in Figure 20.13 (left graph). Precisely, step duration decreases ($p<0.001$) with mechanical efficiency ($n = 5$, $R^2 = 0.9849$, $r = 0.9924$) (right graph);

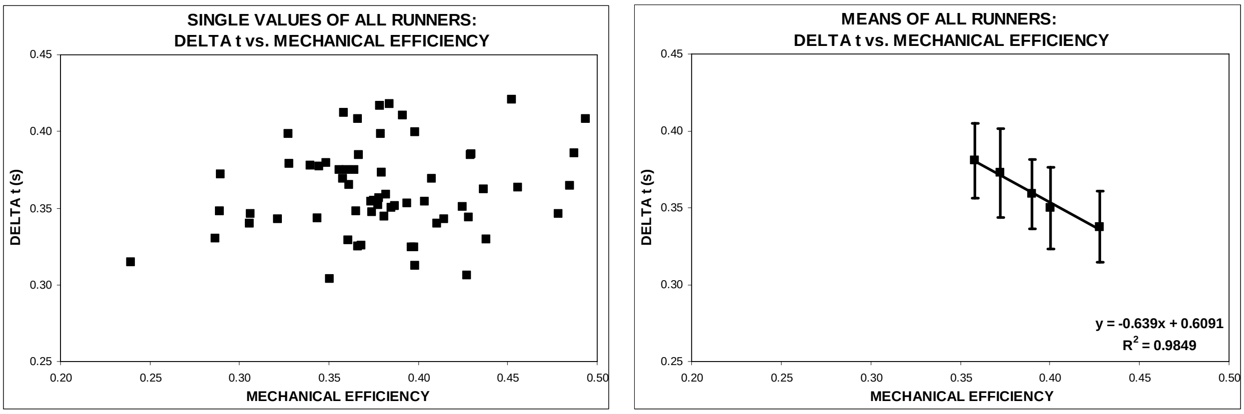


Figure 20.13. Single values of both mechanical efficiency and delta t in all runners (on the left) and delta t means as a function of mechanical efficiency means (on the right).

- differently, no significant differences are found in both forward and vertical directions (n = 5, R^2 = 0.4594, r = 0.6778 and n = 5, R^2 = 0.1429, r = 0.3780, respectively);
- finally, single values of both mechanical efficiency (η) and lateral variability (lateral ΔBCOM) in all runners are graphically represented in Figure 20.14 (left graph). Precisely, ΔBCOM in lateral direction decreases ($p<0.001$) with mechanical efficiency (n = 5, R^2 = 0.9754, r = 0.9876) (right graph).

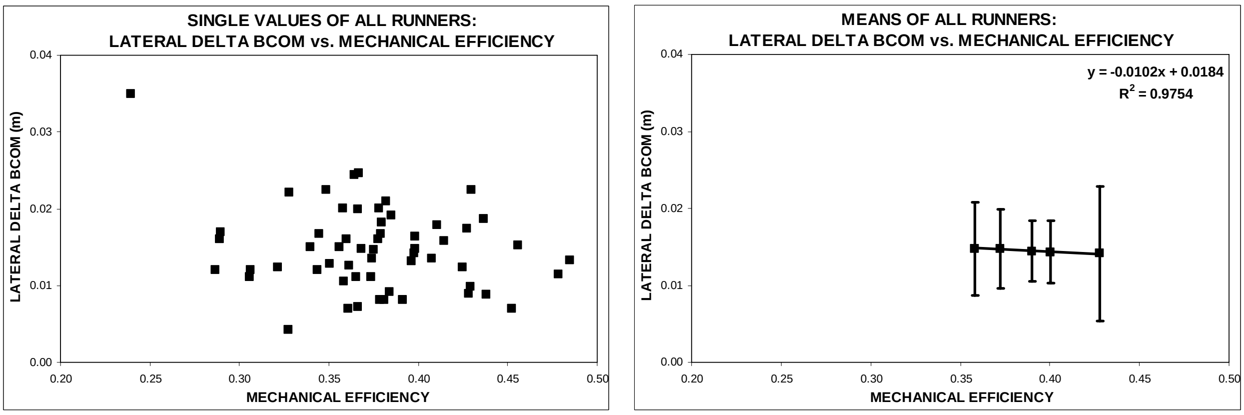


Figure 20.14. Single values of both mechanical efficiency and lateral delta BCOM displacement in all runners (on the left) and lateral delta BCOM means as a function of mechanical efficiency means (on the right).

5.6.2. Relative variability of the body centre of mass *per* speed

A. Coefficients of variation (CV) of both temporal and spatial variability have been calculated as a function of speed, independently of running abilities. All values are reported in Table 20.8:

SPEED (m/s)	CV Δt	CV ΔBCOM Forward direction	CV ΔBCOM Vertical direction	CV ΔBCOM Lateral direction
2.22	5.387 ± 0.816	34.826 ± 10.546	4.666 ± 1.300	47.247 ± 6.868
2.78	5.554 ± 1.415	28.953 ± 4.727	4.522 ± 1.306	50.093 ± 12.430
3.33	5.358 ± 0.887	28.384 ± 7.050	4.568 ± 0.999	53.088 ± 17.333
3.89	5.104 ± 1.343	29.709 ± 6.083	4.791 ± 1.526	44.831 ± 14.382
4.44	4.448 ± 1.279	25.631 ± 7.284	5.517 ± 1.042	40.523 ± 10.827

Table 20.8. Coefficients of variation of variability of the BCOM *per* speed, independently of running abilities.

Precisely, our coefficients of variation show that:

- as illustrated in Figure 20.15 (left graph), step duration (Δt) slightly decreases ($p<0.05$) with running speed ($n = 5$, $R^2 = 0.7157$, $r = 0.8459$);
- as illustrated in Figure 20.15 (right graph), ΔBCOM in forward direction slightly decreases ($p<0.05$) with running speed ($n = 5$, $R^2 = 0.6921$, $r = 0.7157$);
- however, no significant differences are found in both vertical and lateral directions ($n = 5$, $R^2 = 0.5850$, $r = 0.7648$ and $n = 5$, $R^2 = 0.3753$, $r = 0.6126$, respectively).

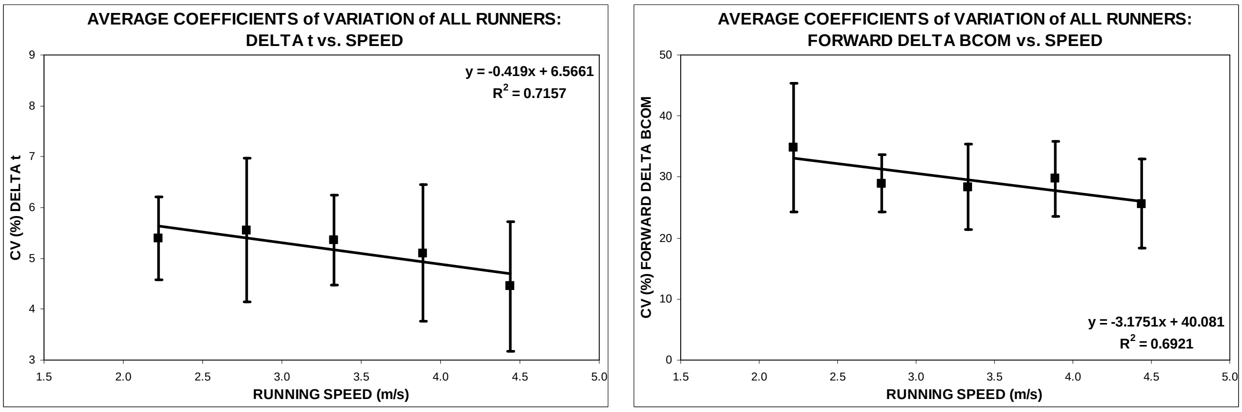


Figure 20.15. Coefficients of variation of delta t (on the left) and forward delta BCOM (on the right) as a function of running speed.

B. If the metabolic cost (its average values are in Table 20.7b above) is considered as a function of the variability of the BCOM, our coefficients of variation show that:

- no significant differences are found in step duration ($n = 5$, $R^2 = 0.3548$, $r = 0.5965$);
- similarly, no significant differences are found in vertical and lateral directions ($n = 5$, $R^2 = 0.2560$, $r = 0.5059$ and $n = 5$, $R^2 = 0.2367$, $r = 0.4865$, respectively);
- single values of both forward variability (forward ΔBCOM) and metabolic cost (C) in all runners are graphically represented in Figure 20.16 (left graph). Precisely, metabolic cost slightly increases ($p<0.05$) with forward displacement of the BCOM ($n = 5$, $R^2 = 0.7945$, $r = 0.8913$) (right graph).

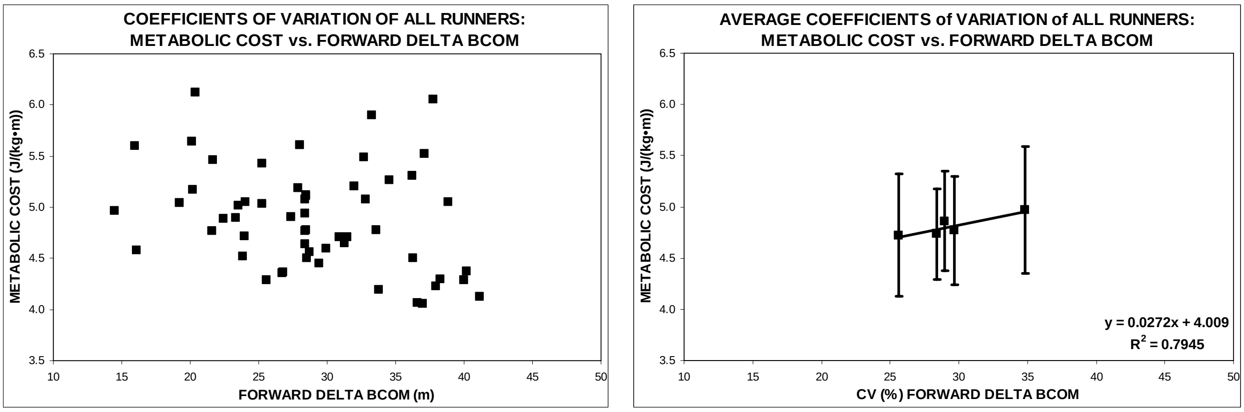


Figure 20.16. Single values of both forward delta BCOM and metabolic cost in all runners (on the left) and metabolic cost as a function of forward delta BCOM coefficients of variation (on the right).

C. If the variability of the BCOM is considered as a function of mechanical 'apparent' efficiency (its average values are in Table 20.7c above), our coefficients of variation show that:

- single values of both mechanical efficiency (η) and step duration (Δt) in all runners are graphically represented in Figure 20.17 (left graph). Precisely, step duration decreases ($p<0.01$) with mechanical efficiency (n = 5, $R^2 = 0.7987$, r = 0.8937) (right graph);

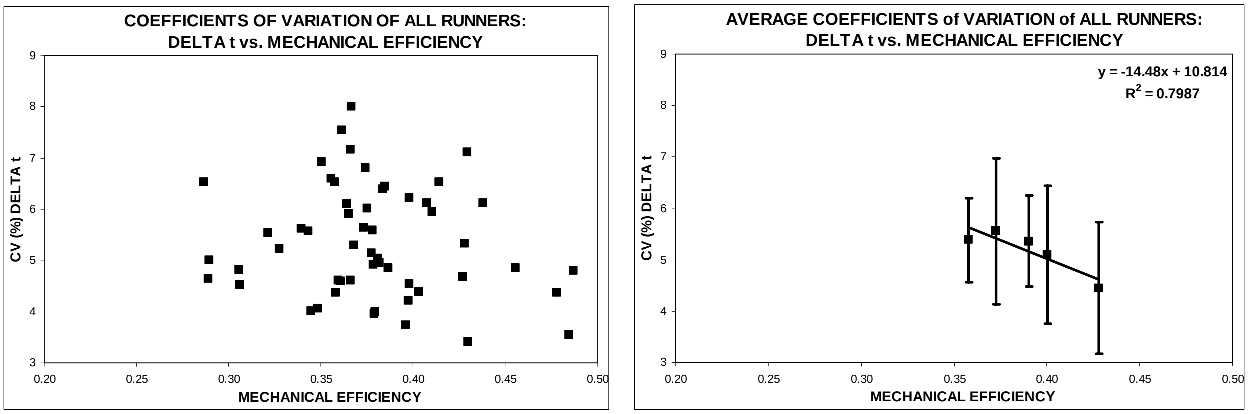


Figure 20.17. Single values of both mechanical efficiency and delta t in all runners (on the left) and delta t coefficients of variation as a function of mechanical efficiency (on the right).

- single values of both mechanical efficiency (η) and forward variability (forward ΔBCOM) in all runners are graphically represented in Figure 20.18 (left graph). Precisely, ΔBCOM in forward direction decreases ($p<0.05$) with running speed (n = 5, $R^2 = 0.7233$, r = 0.8505) (right graph);

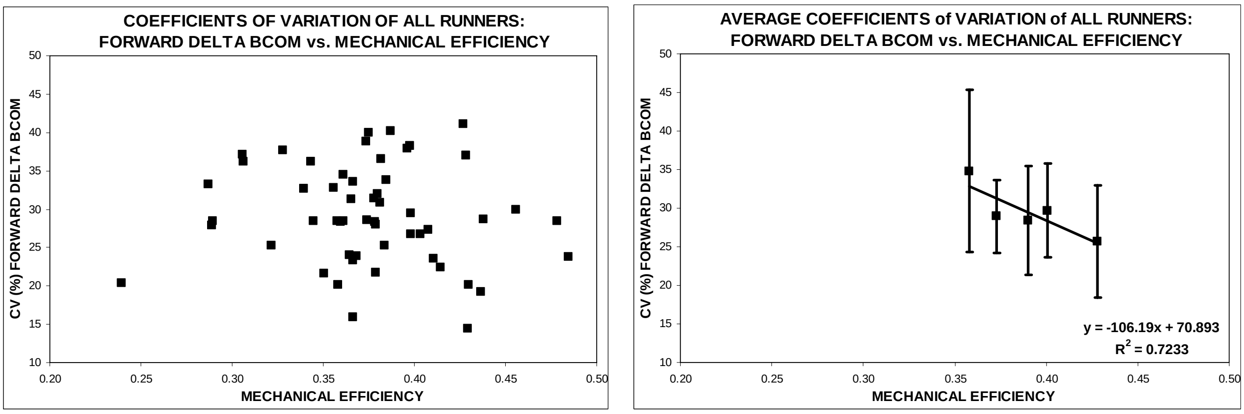


Figure 20.18. Single values of mechanical efficiency and forward delta BCOM in all runners (on the left) and forward delta BCOM coefficients of variation as a function of mechanical efficiency (on the right).

- however, no significant differences are found in vertical and lateral directions (n = 5, R^2 = 0.6873, r = 0.8290 and n = 5, R^2 = 0.4034, r = 0.6351, respectively).

5.6.3. Absolute variability of the body centre of mass *per* subject

Because no significances were found in all the investigated variables (non-paired *t*-tests), we have decided to analyse variability of the BCOM independently of individual running abilities (n = 14).

A. Means (M) of both temporal and spatial variability have been calculated, independently of running speed. All values are reported in Table 20.9a:

RUNNER (R)	M Δt	M ΔBCOM Forward direction	M ΔBCOM Vertical direction	M ΔBCOM Lateral direction
R#1	0.347 ± 0.016	0.035 ± 0.006	0.075 ± 0.008	0.013 ± 0.004
R#2	0.350 ± 0.028	0.018 ± 0.003	0.082 ± 0.010	0.019 ± 0.009
R#3	0.395 ± 0.022	0.017 ± 0.001	0.114 ± 0.007	0.015 ± 0.006
R#4	0.371 ± 0.023	0.019 ± 0.001	0.092 ± 0.005	0.021 ± 0.004
R#5	0.346 ± 0.010	0.019 ± 0.006	0.055 ± 0.004	0.041 ± 0.007
R#6	0.360 ± 0.018	0.017 ± 0.001	0.096 ± 0.004	0.016 ± 0.004
R#7	0.404 ± 0.017	0.016 ± 0.002	0.108 ± 0.006	0.013 ± 0.006
R#8	0.322 ± 0.016	0.013 ± 0.002	0.078 ± 0.003	0.015 ± 0.002
R#9	0.344 ± 0.031	0.016 ± 0.004	0.084 ± 0.012	0.017 ± 0.004
R#10	0.340 ± 0.012	0.017 ± 0.002	0.079 ± 0.004	0.009 ± 0.001
R#11	0.377 ± 0.013	0.013 ± 0.006	0.085 ± 0.009	0.013 ± 0.005
R#12	0.347 ± 0.013	0.014 ± 0.002	0.082 ± 0.005	0.016 ± 0.004
R#13	0.355 ± 0.009	0.012 ± 0.006	0.083 ± 0.008	0.012 ± 0.001
R#14	0.396 ± 0.020	0.005 ± 0.001	0.113 ± 0.014	0.005 ± 0.004

Table 20.9a. Means of variability of the BCOM *per* subject, independently of running speeds.

Moreover, in each subject, average values of metabolic cost (± S.D.), independently of running speed, are presented in Table 20.9b:

RUNNER	METABOLIC COST (J/(kg·m))
R#1	5.214 ± 0.272
R#2	5.460 ± 0.578
R#3	5.384 ± 0.266
R#4	4.898 ± 0.260
R#5	4.554 ± 0.280
R#6	4.861 ± 0.518
R#7	5.330 ± 0.329
R#8	4.323 ± 0.157
R#9	5.107 ± 0.551
R#10	4.632 ± 0.366
R#11	4.915 ± 0.239
R#12	4.255 ± 0.158
R#13	4.078 ± 0.325
R#14	4.071 ± 0.100

Table 20.9b. Average values of metabolic cost, independently of running speeds.

If the metabolic cost is considered as a function of the variability of the BCOM, our means show that:

- no significant differences are found in step duration ($n = 14$, $R^2 = 0.2610$, $r = 0.5109$);
- similarly, no significant differences are found in forward, vertical and lateral directions ($n = 14$, $R^2 = 0.2850$, $r = 0.5338$; $n = 14$, $R^2 = 0.0340$, $r = 0.1844$; $n = 14$, $R^2 = 0.0050$, $r = 0.0707$, respectively).

B. In each subject, average values of mechanical efficiency (± S.D.), independently of running speed, are presented in Table 20.9c:

RUNNER	MECHANICAL EFFICIENCY
R#1	0.310 ± 0.050
R#2	0.320 ± 0.066
R#3	0.396 ± 0.045
R#4	0.384 ± 0.038
R#5	0.408 ± 0.071
R#6	0.451 ± 0.087
R#7	0.393 ± 0.024
R#8	0.393 ± 0.012
R#9	0.364 ± 0.028
R#10	0.397 ± 0.033
R#11	0.385 ± 0.043
R#12	0.382 ± 0.038
R#13	0.374 ± 0.027
R#14	0.478 ± 0.039

Table 20.9c. Average values of mechanical efficiency, independently of running speeds.

If the variability of the BCOM is considered as a function of mechanical efficiency, our means show that:

- no significant differences are found in step duration (n = 14, R^2 = 0.1500, r = 0.3873);
- similarly, no significant differences are found in vertical and lateral directions (n = 14, R^2 = 0.1914, r = 0.4375 and n = 14, R^2 = 0.0009, r = 0.0030, respectively);
- single values of both mechanical efficiency (η) and forward variability (forward ΔBCOM) in all runners are graphically represented in Figure 20.19 (left graph). Precisely, ΔBCOM in forward direction decreases ($p<0.01$) with mechanical efficiency (n = 14, R^2 = 0.4290, r = 0.6550) (right graph).

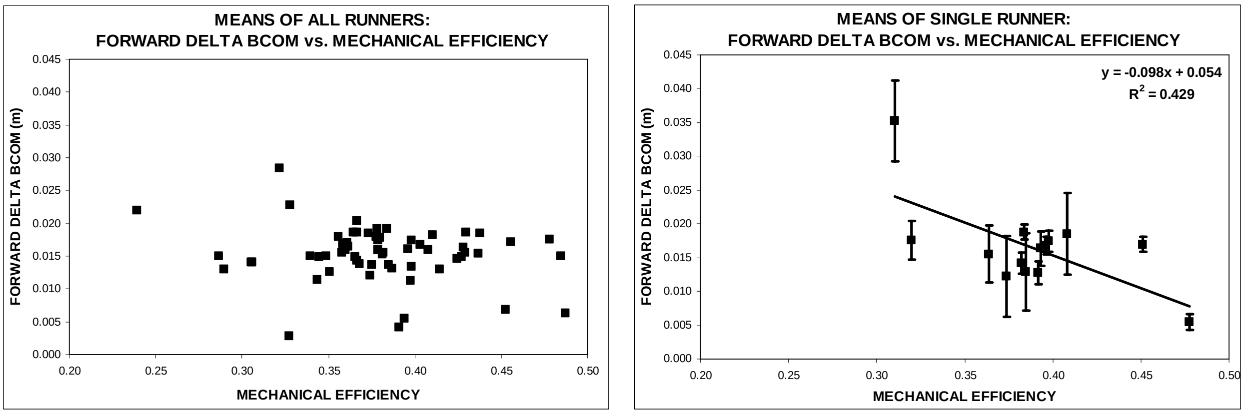


Figure 20.19. Single values of both mechanical efficiency and forward delta BCOM in all runners (on the left) and forward delta BCOM means as a function of mechanical efficiency means (on the right).

5.6.4. Relative variability of the body centre of mass *per* subject

A. Coefficients of variation (CV) of both temporal and spatial variability have been calculated, independently of running speed. All values are reported in Table 20.10:

RUNNER	CV Δt	CV ΔBCOM Forward direction	CV ΔBCOM Vertical direction	CV ΔBCOM Lateral direction
R#1	4.716 ± 0.194	32.857 ± 4.367	5.314 ± 0.791	54.260 ± 9.345
R#2	5.975 ± 0.710	30.129 ± 5.502	5.543 ± 0.731	56.014 ± 15.826
R#3	3.929 ± 0.771	21.852 ± 3.550	3.280 ± 0.302	32.013 ± 7.241
R#4	6.374 ± 0.948	25.748 ± 2.919	5.973 ± 0.595	44.339 ± 7.591
R#5	4.298 ± 1.156	27.073 ± 9.937	3.617 ± 1.610	28.572 ± 14.301
R#6	4.137 ± 0.532	34.944 ± 13.562	4.950 ± 0.483	49.467 ± 8.188
R#7	6.893 ± 0.428	18.551 ± 5.834	3.140 ± 0.987	44.759 ± 3.799
R#8	4.750 ± 0.576	35.607 ± 6.211	4.956 ± 1.706	51.103 ± 10.139
R#9	6.104 ± 0.786	26.849 ± 6.647	5.204 ± 0.656	51.282 ± 8.087
R#10	5.349 ± 0.572	33.808 ± 4.183	4.096 ± 0.548	61.459 ± 13.602
R#11	4.573 ± 1.123	27.003 ± 2.182	4.184 ± 0.762	45.284 ± 3.112
R#12	5.595 ± 1.246	35.379 ± 4.435	5.973 ± 1.081	43.182 ± 5.349
R#13	6.728 ± 1.146	29.857 ± 2.040	5.642 ± 0.016	68.341 ± 1.022

Table 20.10. Coefficients of variation *per* subject, independently of running speeds.

If the metabolic cost (its average values are in Table 20.9b above) is considered as a function of the variability of the BCOM, our coefficients of variation show that:

- no significant differences are found in step duration (n = 14, R^2 = 0.0033, r = 0.0574);
- similarly, no significant differences are found in vertical and lateral directions (n = 14, R^2 = 0.0859, r = 0.2931 and n = 14, R^2 = 0.0134, r = 0.1157, respectively);
- single values of both forward variability (forward ΔBCOM) and metabolic cost (C) in all runners are graphically represented in Figure 20.20 (left graph). Precisely, metabolic cost slightly decreases ($p<0.05$) with forward displacement of the BCOM (n = 14, R^2 = 0.3603, r = 0.6002) (right graph).

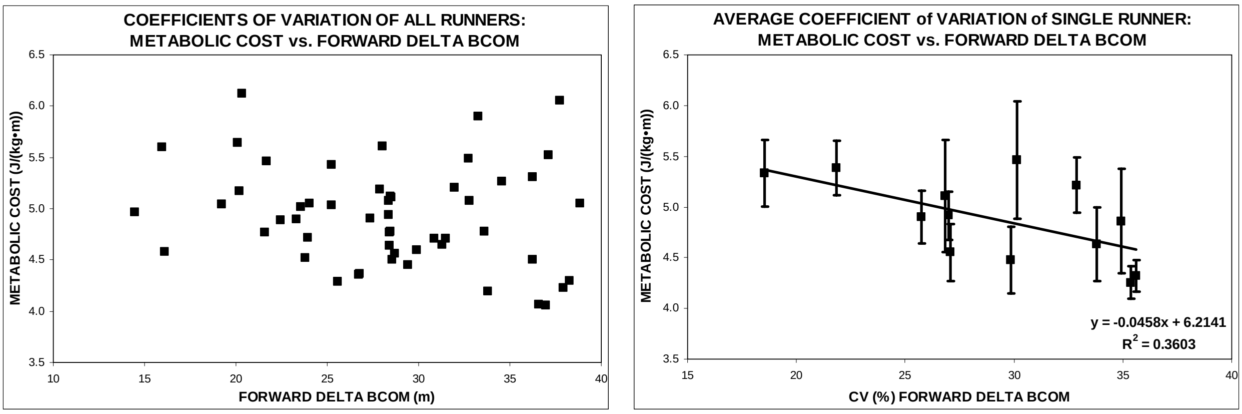


Figure 20.20. Single values of both forward delta BCOM and metabolic cost in all runners (on the left) and metabolic cost as a function of forward delta BCOM coefficients of variation (on the right).

B. If the variability of the BCOM is considered as a function of mechanical efficiency (its average values are in Table 20.9c above), our coefficients of variation show that:

- no significant differences are found in step duration (n = 14, R^2 = 0.1020, r = 0.3194);
- furthermore, no significant differences are found in spatial 3D variability of the BCOM (n = 14, R^2 = 0.0000, r = 0.0000 in forward direction; n = 14, R^2 = 0.1540, r = 0.3924 in vertical direction; and n = 14, R^2 = 0.1220, r = 0.3493 in lateral direction).

6. CONCLUSION

To sum up, our biomechanical analysis permits us to conclude that:

- stride frequency and stride length slightly increases as a function of running speed, independently of running ability. Moreover, stride frequency seems to be greater in more trained subjects; however stride length seems to be smaller in these runners;
- duty factor decreases as a function of running speed, independently of running ability. It seems to be higher in more trained runners;

- mechanical external work does not change with speeds; internal work slightly increases with speed, independently of running ability. Only slightly differences are found among different running abilities;
- mechanical external work calculated by using the continuous function (Fourier analysis) is significantly smaller than external work measured discretely (cycle by cycle);
- mechanical total work does not change with speeds. Only slightly differences are found among different running abilities;
- more trained runners are more efficient than no-trained runners;
- in both absolute and relative temporal and spatial variability of the BCOM, slightly significances are found: precisely, **a)** absolute step duration and lateral displacement of the BCOM decrease with speed; **b)** relative step duration and forward displacement of the BCOM decrease with speed; **c)** absolute temporal and spatial variability (i.e. forward and lateral directions) of the BCOM change slightly in relation with metabolic cost; **d)** relative spatial variability (i.e. forward direction) of the BCOM changes slightly in relation with metabolic cost; **e)** the higher the metabolic cost, the higher is the corresponding vertical displacement of the BCOM (with increasing energy consumption); **f)** the smaller the step duration, the higher is the metabolic cost. Therefore, these conclusions slightly confirm the initial hypothesis that the increases in metabolic cost could be related both to three-dimensional displacements of the BCOM and the stride frequencies; **g)** absolute temporal and spatial variability (i.e. lateral direction) of the BCOM changes in relation with mechanical 'apparent' efficiency; and **h)** relative temporal and spatial variability (i.e. forward direction) of the BCOM changes in relation with mechanical efficiency.

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PART 3

CONCLUSIONS

Chapter 21

CONCLUSIONS

1. INTRODUCTION

In literature, it has been widely demonstrated the high importance of estimating symmetries (in its various expressions) in order to select the best individual (human or animal) performer and environment (Manning et al., 1994; 1998). Clearly, this topic might be critical both in healthy (e.g. daily life) and improved (e.g. competitions and ratings) contexts. However, there is evidence that a correlation between static anatomical symmetries and running economy exists only in animals.

Therefore, the finding of such a relationship in humans constitutes the obvious aim of this study. At the same time, this purpose represents its major novelty, too. As already tested in animals, it makes possible to test for (static anatomical and dynamic functional) symmetries detection runners featuring different abilities.

The repeatability of such a test protocol in different moving conditions (e.g. cycling) could help in better characterizing and qualifying the effects of different training programs.

In the next sections, we will briefly discuss main peculiarities, limits and future developments of our research project. However, in depth, at the end of each chapter single results have been widely discussed and commented according to previous references.

2. MAIN PECULIARITIES

In literature, this research is the first effort to study and solve symmetries and running performance in humans. As discussed about horse performances (Manning et al., 1994), selection and investigation of runners with different training ability represent a crucial point. Indeed, we try to verify if (static and dynamic) symmetries represent a discriminating factor which could satisfactorily explain differences firstly a) in physiological parameters (i.e. metabolic cost) and, finally, b) in running performance in various level abilities.

All our measurements (i.e. recordings of kinematic and physiological data) have been made by the same subject thus avoiding operator-dependent errors. Furthermore, the same climatic conditions (i.e. humidity and temperature) have been kept.

Similar anthropometric dimensions (i.e. height and body mass) allow sample groups, which are very homogeneous and effectively comparable.

The recording of Magnetic Resonance Imaging makes up a qualitative and quantitative value added. Indeed, MRI is one of the most valid and accurate tools to assess static anatomical symmetries (Raines, 1972; Deck et al., 1989; Heymsfield et al., 1997; Brown et al., 1999). Positively, we have demonstrated that it is possible to investigate the main anatomical regions (i.e. inferior limbs) by using this simple and easily-available approach. In addition, a strong relationship between ankle and knee areas has been highlighted in all runners. Moreover, the mathematical application of different methods to analyse both static and dynamic symmetries plays an important role in developing techniques and systems, which try to analyse resulting images in an innovative way. The implementation of these programs in LabVIEW environment constitutes an important step for further researchers who want to develop and apply the same approaches.

As previously demonstrated in the first study, both the Digital Locomotory Signature and the Symmetry Index seem to be a good mathematical and graphical solution to describe the 3D BCOM displacement along each movement direction. In effect, independently of training ability, the BCOM raises and lifts slightly as a function of speed. Furthermore, right and left steps are mostly asymmetrical in the forward direction and symmetrical in the vertical. This last result is probably due to the combined action of gravity force and its opposite ground reaction force. In addition, differently to what was expected, slight differences have been found among runners. On the whole, the globally asymmetry is probably related both to anatomy (i.e. leg length) and which hand the studied people use (i.e. predominant right handedness).

The comparison between over-ground and treadmill locomotion wholly investigates physiological differences among these surfaces. As a result, our data can concur (and complete) with existing literature, although their assessment has shown no significant differences.

Furthermore, the knowledge of the main simple and complex biomechanical variables constitutes another significant value added. Indeed, literature concerning the pattern of such parameters at different running conditions is not fully comprehensive and thorough. Thus, our results can fulfil the research outlook.

Finally, the study of the variability (spatial and temporal) of the BCOM wholly complete and extend the previous documented data, although slight relationships have been highlighted.

3. MAIN LIMITATIONS (OR DISADVANTAGES)

First of all, it is important to underline the absence of significant differences in physiological parameters (e.g. metabolic cost) as level running abilities change. Indeed, the slight differences which were found in both static and dynamic symmetries were not supported by running economy differences. This will be partially in contrast to our initial (implicit) hypothesis and to the existing literature. To solve this problem, in the near future, it will be fundamental:

a) to maintain a single speed longer to better isolate and investigate physiological (and kinematics) adjustments. For instance, in such a way, metabolic cost could be recorded after 1 hour running;
b) to use another appropriate device (i.e. Quarkb^2 or Douglas bags) to record bioenergetics of running, as well;
c) to carry out experiments on treadmill with a different stiffness;
d) to increase the number of anatomical regions investigated with MRI and then compared to running economy (e.g. superior limbs and human face).

In addition, the restricted number of subjects we investigated represents an important disadvantage. Indeed, in some circumstances, negatively, the statistical analysis has been applied only for occasional and skilled runners. Clearly, more subjects could a) evidence higher differences both in symmetries and running economy and b) complete these statistical analyses.

Furthermore, only few runners were able to carry out all the test protocol. As a consequence, results which were obtained at the highest speeds refer to just a limited group. Among the others, the wide range in age among runners could partially represent a) a disadvantage and b) a possible explanation for those subjects who did not complete all the protocol.

To be precise, the bi-dimensional approach that has been used to study static symmetries to assess the main differences among groups seems not to be so adequate and valid. This important conclusion depends on the poor significance that has been found among groups independently of the anatomical region. Probably, it is due to a partial (not complete) analysis of single Magnetic Resonances Images. In other words, it will be urgent to analyse them by using other software (e.g. Osiris), which shows in depth every anatomical area in all planes. In this way, it will be possible to analyse the pelvic region, as well. Indeed, this area could not be investigated because of the chosen MRI protocol.

Furthermore, the three-dimensional approach has to be optimized in order to calculate a cross-correlation coefficient separately for each anatomical region/segment. It might be necessary to include in such analysis the hip joint, too.

As far as right/left symmetry and biomechanical variables have been concerned, only kinematic data has been recorded. However, the knowledge of both kinetics and electromyography seems to be so important in recording the main joint angles, forces and moments, and the activity of muscles in the lower limbs. Only slight significant differences have been found in both 3D *contours* and Symmetry Indexes. Thus, quite similar patterns could be found independent of running ability (Minetti, 2006).

4. FUTURE DEVELOPMENTS

In the near future, we will study a larger number of more trained runners (i.e. belonging to top runner group) in order to fill up the studied samples. Moreover, each runner will be tested at the same speeds maintained for a longer time. This will be achieved to investigate the probable adaptation of the physiological and kinematics parameters as running time increases.

Finally, it will be possible to use different physiological (i.e. NIRS, Portapres, Innocor and Accusport) and biomechanical (i.e. force platforms and EMG) equipment to record both a) cardiac output, b) arterial pressure profile, c) lactate concentration, d) joint angles, forces and moments, and e) the muscular activity of main muscles in the lower limbs and trunk. In such a way, a more complete characterization of each individual could be drawn up.

The bi-dimensional method, proposed to analyse static symmetries, will be improved by using different software (e.g. Osiris) and analysing the pelvic region. Moreover, to complete this analysis main static digital high-resolution pictures (frontal, lateral side and posterior) could be recorded and investigated. Therefore, the three-dimensional method will be modified in order to compare better the main anatomical areas among different trained groups.

Importantly, as previously demonstrated in the first study, the mathematical method (i.e. Fourier Series) and the valid evaluation protocol for the study of 3D BCOM pattern (and its symmetry) we proposed have worked satisfactory.

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