Adaptability of Human Stepping Patterns

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WIEBREN ZIJLSTRA

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RIJKSUNIVERSITEIT GRONINGEN

ADAPTABILITY OF HUMAN STEPPING PATTERNS

Proefschrift

ter verkrijging van het doctoraat in de Medische Wetenschappen
aan de Rijksuniversiteit Groningen
op gezag van de
Rector Magnificus, Dr. F. van der Woude,
in het openbaar te verdedigen op
woensdag 15 oktober 1997
des namiddags te 4.15 uur

door

Wiebren Zijlstra

geboren op 8 juli 1963
te Utrecht

Stellingen

bij het proefschrift
Adaptability of human stepping patterns

van Wiebren Zijlstra

1 A paradigm is what the members of a scientific community share, and conversely, a scientific community consists of men who share a paradigm (Kuhn).

2 Gemeinsame enge Horizonte verschaffen das Gefühl von gesamthafter Weitsicht (Jaspers).

3 De studie van stoornissen in het menselijk bewegen vereist een benadering die naast diepgaande vakspecifieke kennis een weidse blik op andere vakgebieden toestaat.

4 Het nemen van klinische beslissingen op basis van een vergelijking met normaalwaarden is met betrekking tot bewegingsanalyses van loop- en balansstoornissen tot op heden niet mogelijk.

5 "Because a wide variety of stride length-stride frequency combinations can occur, these two dimensions should always be measured and specified whenever data on gait are presented" (Inman et al. 1981, dit proefschrift).

6 "..... research into quiet standing is very limited in revealing the mechanisms of balance and as a diagnostic tool to pinpoint deficits of the system" (Winter, Gait & Posture 1995; 3: 200).

7 Te gemakkelijk wordt in vele studies van het menselijk bewegen voorbijgegaan aan verschillen tussen proefpersonen en verschillen tussen herhaald uitgevoerde bewegingen.

8 Al wordt het verrichten van toegepast onderzoek nagestreefd, de praktijk en theorie van bewegingsanalyse verdienen in elke studie Bewegingswetenschappen een fundamentele plek.

9 Zolang de biomechanica van een complexe beweging onvoldoende is begrepen, is de kans groot dat nadenken over de "centrale" sturing van deze beweging te vroeg komt.

10 Voor het legitimeren van (para)medische behandelingen van houdings- en bewegingsstoornissen zijn eenvoudige meetinstrumenten die ziektespecifieke veranderingen in motorisch functioneren kunnen vastleggen onmisbaar.

11 De relatie tussen technologische complexiteit van bewegings-analyses en klinische relevantie is vooralsnog onbekend.

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Chapter 1: General Introduction

Aims and outline of thesis

In this thesis a number of studies are presented which explore the characteristics of human stepping patterns during walking in different conditions. The studies were performed in different groups of subjects, i.e. children, healthy young adults, healthy elderly subjects, and patients with Parkinson's disease (PD). The studies were performed in order to identify characteristic differences in stepping patterns and, when possible, to explain these differences. The main motivation for these studies are the gait disturbances which can be observed in neurological patients, particularly patients with Parkinson's disease, and in elderly people without a clearly defined pathology. Although gait disturbances are often a characteristic feature of these groups, the deviating stepping patterns are not always well documented, and they are often misinterpreted. Also, stepping patterns in children are different from the patterns observed in healthy adult subjects, a characteristic often attributed to an immature central nervous system (CNS). The role of the CNS in determining the nature of both normal and abnormal stepping patterns is unclear. This lack of knowledge underlies the problems in interpreting characteristic changes in stepping patterns in elderly with or without a clearly defined pathology. Therefore, a specific aim of this thesis is to document variant and invariant aspects of stepping patterns in different conditions, and to investigate which factors explain these aspects. The general purpose of this thesis is to contribute to the clinical analysis and interpretation of deviations in stepping patterns of a presumed central origin.

The remainder of this introductory chapter aims to give a perspective on movement coordination and clinical gait analysis, and to explain the motivation for performing the studies presented in this thesis. First, relevant literature with respect to theories about the coordination of human movement will be discussed briefly. Thereafter, subsequent sections will discuss the following: control aspects of posture, gait, and equilibrium; motor disturbances in PD; clinical analysis of human movement and gait; and lastly, disturbances of gait and balance in elderly. The introductory chapter ends with a section which shortly describes the work presented in this thesis. Subsequent chapters will present the results of different studies, and a final chapter contains a general discussion of these studies.

Coordination of human movement

Starting about a century ago with the work of Muybridge, Marey, Braune & Fischer and others[1], the possibility to objectively describe complex human and animal movements in time have led to a present wealth of knowledge about regularities in movement performance. Although biomechanical analyses did much to help us understand these regularities, our insight into the role of the CNS in movement coordination remains rather fragmented. A complete understanding of movement coordination requires knowledge from all parts of the neuro-musculoskeletal system, the way these parts interact, and the interaction of this system with its environment. Thus, the study of human movement involves a diversity of scientific disciplines which all study different aspects of movement production, and which often have different theoretical and experimental approaches.

One approach to the study of human movement is biomechanics, here simply defined as the application of Newton's laws of motion to the behaviour of biological organisms. Biomechanical analyses have proven to be very useful in the study of locomotion in different animal species[2-4]. Many regularities in human movement performance can also be understood by means of concepts developed from biomechanical analyses. For example, several aspects of the observed kinematic patterns during locomotion, or during the maintenance of standing posture, can be understood by means of inverted pendulum models of the human body[5,6]. Inverse dynamics and modelling approaches to movement are the most powerful tools of biomechanics for the analysis of human movement. Inverse dynamics refers to a method in which kinetic quantities (forces, moments) are derived from a kinematic description of a movement. The method requires modelling of the movement by means of a linked-segment model, a kinematic description, and, in case of forces being exerted on the moving body, the measurement of contact forces. Inverse dynamics offers the possibility to analyse the way forces, or moments, are exerted during movements. Thus, it may help to understand the way muscular activity contributes to the generation of force, and also in estimating the amount of stress on tissues. Another example of the contribution of biomechanics to understanding aspects of movement can be found in analyses of the mechanical work during locomotion. Such analyses have shown that during walking, a recovery of mechanical energy exists[7], and that a minimum energy expenditure per unit distance is achieved at a self-selected speed[8]. Furthermore, metabolic analyses have shown that stepping patterns at different speeds are coordinated in such a way that the observed spontaneous movement patterns require the least energy expenditure[9,10]. The examples given above illustrate that biomechanics can reveal important aspects of the coordination of movement: i.e. the forces which underlie our movements, the way we stabilise posture, or maintain equilibrium; the way stress on our musculoskeletal system is kept within certain limits; and the way energy consumption is kept to a minimum by so called "optimal" coordination. Hence, biomechanics can explain regularities in movement performance which are a consequence of the constraints to which the human body is subjected as a physical system interacting with a physical environment. Theories or hypotheses about the role of the CNS in motor behaviour should therefore include knowledge of mechanical aspects of movement production. When possible, they should dissociate the effects of activity of the CNS (active factors) and those of passive mechanical factors in determining the characteristics of movements.

Independent from developments in biomechanics, several other disciplines which study movements in biological systems have contributed to an increased knowledge of movement coordination and brain activity. Based upon correlative studies involving neuroanatomy, neurophysiological activity, and motor functions, many attempts have been made to attribute specific roles in movement coordination to neuroanatomical structures like the spinal cord, cerebellum, different cortical areas, and basal ganglia[11]. In complex human movements, the different central control processes and their interaction cannot easily be identified. Therefore, much of our current knowledge stems from animal experimental work. This work includes experiments with different (in)vertebrate species in which remaining motor functions are assessed after the nervous system is severed at a particular level and animal experimental studies of the neural activity in different brain regions during, or preceding, movement execution. Observations in humans often rely on studies using ingenious methods, studies of different tasks and conditions, and observations of motor disturbances in patients with specific central lesions. Studies of this kind may involve experiments in which activity in different brain regions is assessed using techniques as electroencephalography, radioisotope scanning, magnetoencephalography, and, lately, positron emission tomography in combination with an assessment of motor behaviour.

In neurophysiology and in psychology, similar concepts were developed about the way the CNS contributes to the coordination, or control, of movement. In neurophysiological research, both the nature of reflex responses in a variety of motor tasks, and the ability of the CNS to generate patterns of neural activity in absence of feedback, have been extensively studied (for a review see Ref. 12). From animal experimental studies of rhythmic movements, the central pattern generator concept was developed as a basis for understanding the ability of the CNS to autonomously generate rhythmic activity[13,14]. In psychology, several engineering oriented approaches to movement behaviour have lead to theories about motor learning and motor control. In the early attempts to define the role of the CNS in the control of movement, two major diametrically opposed approaches could be discerned; 1) a centralist view which assumes that a movement is controlled by a central representation of the movement called "motor program". This central program is held responsible for the "open loop" execution of certain classes of movement which do not require conscious monitoring, or feedback[15]. 2) a peripheralist view which assumes that all movements are executed on a feedback basis of control. This "closed loop" theory of movement control assumes that movement related sensory feedback is used for detecting errors in movement execution and for the subsequent correction of the ongoing movement[16]. To this purpose different central representations are assumed for selecting and initiating movements, and for comparing actual feedback with expected feedback.

As the debate between these two extreme positions seemed to be caused by focusing on entirely different aspects of movement behaviour, the Schema Theory was proposed as a general theory of motor learning and motor control which encompasses both "open" and "closed loop" mechanisms[17]. The Schema Theory assumes different central representations, and it postulates the existence of generalised motor programs which can execute a movement with all the details of the movement determined in advance. A motor program can be initiated after a number of computational phases in which response selection and parametrisation takes place. The generalised motor program has variant and invariant characteristics[17,18]. Examples of variant characteristics are speed of movement execution, required force, movement amplitude and movement direction. Relative timing of a movement pattern is considered to be an invariant characteristic. Regardless of differences in speed, direction, or movement context, the motor program always generates a movement on an invariant (relative) time scale.

The above mentioned physiological and psychological approaches to movement coordination all assume central representations of motor behaviour (albeit that the neural basis for the assumed processes and representations is not always clear), and they are characterised by a computational account of motor behaviour. This approach has been called the "motor-approach" of movement behaviour[19]. Essential characteristics of the motor approach include: an information processing perspective on the CNS; the assumption that learned tasks are produced by motor plans which are made up of several motor programs; and the attribution of specific roles to neuroanatomical structures, and neurophysiological processes, in the process of programming and executing movements[11]. Within a motor plan, motor programs encode muscle activity for specific subtasks, and they also act as commands for the initiation of other programs. Feedback is supposed to adjust programmed movements.

In the last decades an approach to movement behaviour has emerged which defies the necessity of assuming central representations and computational phases for explaining regularities in movement behaviour[19]. This approach can be considered an elaboration of the work of Bernstein and Gibson who independently developed their ideas on human motor behaviour in the early forties and fifties (representative samples of their work can be found in Refs. 20 and 21, respectively). A frequently used name to delineate this approach is dynamical systems theory, outlines of this approach can be found in Refs. 22-24. Essential to dynamical systems theory is its focus on spatio-temporal patterns of organisation in complex physical and biological systems, and the assumption that organised patterns (attractor states) in systems of high complexity can emerge spontaneously as a result of self-organisation. Dynamical systems theory has shown that non-linear changes in patterns of movement can be well described by changes in so called order parameters. An example of such an order parameter is the relative phase between two limbs or between joints within a limb. Non-linear changes in the behaviour of a complex system, i.e. transitions from one attractor state to another, can be induced by changing certain control parameters. In quadrupedal locomotion, the switch from an alternating to a non-alternating gait pattern as a result of increases in stride frequency is an example of such a transition. Transitions from one pattern to another coincide with changes in the stability of coordinative patterns. Therefore, transitions are characterised by a temporary loss of stability (of the old pattern) until an increased stability is found in the new pattern. Similar dynamic behaviour has been demonstrated in a large range of different phenomena[23]. The dynamical systems theory is able to demonstrate common coordinative principles, not only in different motor behaviour, but also in other coordinative phenomena.

Different approaches to the study of human and animal movement are characterised by differences in all aspects of their respective research programmes. The domains of interest vary from easily observable spatio-temporal characteristics of movements to properties of different parts of the human neuromusculoskeletal system which can only be accessed by means of sophisticated techniques. The central problems and core ideas of the different research programmes vary according to the different domains. The different theories or hypotheses may be directed at explaining phenomena at a cellular level, an organic level, or they may include assumed psychological entities within the CNS which underlie characteristic features of the behaviour of the whole organism. Differences in domain and theory are reflected in a large variety of methodological approaches of studying movement. Conflicts between different theories seem to be centred around the way assumptions are made about the presence or absence of central representations of movement behaviour; what exactly is represented, how is this represented, and where is it represented?

In the work presented in this thesis, it will be assumed that voluntary movement requires initiation and execution of movement by neural activity based on a central representation, and that based on observations of motor behaviour, assumptions about central control processes can be made. However, as the movement patterns resulting from voluntary activity will be the result of an interaction between voluntary and involuntary mechanisms, the characteristics of these movement patterns will to a large part be determined by both passive and active aspects. In the studies presented in this thesis it will be attempted to dissociate passive from active mechanisms in order to identify central control mechanisms in the coordination of stepping movements.

Control aspects of posture, gait, and equilibrium

The central control of voluntary complex movements requires the simultaneous control of different aspects of movement. This simultaneous control requires an interaction of different central control processes which depend on different neuroanatomical structures and neurophysiological processes[11,12,25]. Regardless of the central structures and processes involved, two different aspects can be recognised with respect to the goals of central control of movement[25]: first, the maintenance of certain reference values against internal or external perturbations; and secondly, the coordination of the displacement of body segments, or the whole body, along a certain trajectory towards a specific goal. The maintenance of posture requires that the positions of certain parts of the body, or whole body posture, are stabilised against disturbances. Thus, postural control belongs to the first type of central control. The second type of control relates to the activity which is necessary to initiate and execute the movement itself. In locomotion, this type of control would be involved in the generation of propulsive forces which lead to a desired displacement of the body. However, in most movements, and particularly in locomotion, both types of control have to be combined. This is very obvious when the demands for equilibrium control in different conditions are considered.

During static conditions, such as sitting or quiet standing, equilibrium control, in an indirect way, requires the maintenance of a reference value, i.e. the stabilisation of the projection of the body's centre of mass with reference to the support area. However, this stabilisation is not achieved by postural control only. During quiet standing, an active control of the centre of pressure position relative to the centre of mass projection within the base of support is used for controlling equilibrium in the sagittal plane[6]. This control strategy has been called the ankle strategy[26]. The control processes in these situations seem to be based on slow corrections which are based on continuous feedback of all three major sensory systems (i.e. visual, proprioceptive, and vestibular sensory systems). Kinematic behaviour in this condition can be modelled well with an inverted pendulum model[6]. When quiet standing is disturbed by voluntary actions, or external perturbations, stereotypical patterns of movement and muscle activity can be observed. External perturbations of standing posture result in triggered postural responses which consist of reflex activity, and subsequent voluntary activity. Similar responses may be triggered by different sensory systems. The nature of postural responses may vary according to initial posture, perturbation characteristics, support conditions, and psychological factors such as instruction, expectation, previous experience, and learning. The reflex part of postural responses may differ in timing and amplitude, and much research has focused on the contribution of reflex responses to functional movement behaviour in different conditions[12]. Perturbations of standing can be largely compensated by means of the ankle strategy. However, when this strategy is insufficient, or when the support surface does not allow an effective control of contact forces, trunk movements (i.e. a hip strategy[26]), arm movements, stepping movements, or a mix of these strategies may be observed.

Posture and equilibrium can also be disturbed by voluntary movements. To prevent non-intentional changes of posture and/or centre of mass position, anticipatory postural activity precedes voluntary movements. Anticipatory postural activity acts in a feed-forward and open-loop character to compensate for the expected disturbances in posture and equilibrium due to voluntary initiated movements[25]. The nature of anticipatory activity varies with characteristics of the voluntary movement. For example, the duration of the anticipatory phase depends on load and movement velocity, i.e. on the perturbing effect of the movement[27]. The disturbance of posture or equilibrium by a combination of external and internal causes, for example, perturbation of gait initiation[28,29], results in an interaction of reactive postural activity with anticipatory postural activity and movement related activity.

It has been shown that during locomotion reflexes[30,31], postural responses[32], and the anticipatory postural activity associated with voluntary arm movements[33] all show a phase dependency during the step cycle. This phase dependency reflects the constantly changing conditions for the control of posture, movement, and equilibrium during the stride cycle. During walking, equilibrium control requires stabilisation of the body's centre of mass trajectory with reference to a continuously changing support configuration. Contrary to standing, equilibrium control during walking is not directed at maintaining the centre of mass projection within the base of support[5,6]. Only during the short periods of double support the centre of mass projection remains within the base of support, during the major part of the stride cycle (close to 80%) there is an unstable situation. The centre of mass trajectory can only be controlled by subsequent foot placements (and the resulting ground reaction forces), and an accurate control of the displacement of head, arms, and trunk[6]. The latter is achieved by stabilising an almost erect position of the trunk in the frontal and sagittal planes, and an attenuation of accelerations of the head. The stabilisation of the trunk, and particularly the head, allows a good orientation to the environment by visual and vestibular sensory systems[34].

Thus, dynamic equilibrium is maintained by both postural control (particularly of the head and trunk), and the control of stepping movements from cycle to cycle.

From the above discussed examples of equilibrium control in different conditions, it follows that equilibrium control results from an interaction between postural and movement control. In complex movements, these two cannot be dissociated. The stabilisation of the centre of mass projection during quiet standing, or during external perturbations of standing, is achieved by postural control, active control of the centre of pressure, and, when necessary, compensatory movements (e.g. arm or trunk movements). When threatening to disturb posture, or equilibrium, voluntary movements are preceded by anticipatory postural activity. During locomotion, equilibrium control is achieved by a combination of postural control of head and trunk, and control of foot placements during subsequent steps. Consequently, the spatio-temporal characteristics of stepping movements and ground reaction forces not only reflect propulsive control, but also equilibrium control.

Parkinson's disease

Parkinson's disease (PD) is the most frequently observed progressive neurodegenerative disease in the human population over fifty years of age[35]. Its prevalence in this age group is approximately 1%. Although the precise causes of PD are unclear, the association of Parkinsonian symptoms with a progressive loss of dopaminergic cells in the substantia nigra (a neuroanatomical structure in the mesencephalon (midbrain) is well established. The progressive loss of substantia nigra neurones leads to a dysfunction of the basal ganglia, a group of neuroanatomically and functionally related subcortical nuclei. The basal ganglia dysfunction expresses itself for the clinical observer as a loss of motor function. Although this loss of function predominantly concerns motor behaviour, cognitive and emotive functioning can also be involved. Motor symptoms in PD include problems in initiating movements (akinesia), a reduction (hypokinesia) or slowing down (bradykinesia) of movements, tremor, and rigidity. In later stages of the disease, disturbances in postural and equilibrium control may develop. Diagnosing PD is a matter of the clinician. There is no commonly used laboratory test which accurately establishes the presence of PD. Therefore, the diagnosis of PD is normally accepted when the following criteria are fulfilled: presence over one year or more of two of the cardinal motor signs of PD (bradykinesia, tremor, rigidity) and a lasting responsiveness to treatment with anti-Parkinsonian drugs[36]. In PD, drug therapy primarily consists of maintaining intra-cerebral levels of dopamine by Levodopa or dopamine agonists.

In the last decades, many aspects of the underlying pathology and the treatment of PD have been subjected to research. One major pharmacological development is that current research suggests the possibility of neuroprotective therapies that may be able to put the disease progression to a halt, or at least slow it down[37]. It has been estimated that at least 80% of the dopaminergic substantia nigra neurones are lost before the cardinal signs and symptoms of PD appear. Before this moment, aspecific symptoms may be present but these symptoms do not necessarily direct the clinician to the diagnosis of PD. Because of the presumed future possibilities for neuroprotective medication, an early diagnosis of PD is essential. An early diagnosis requires the availability of reliable and sensitive test procedures[38]. Other developments in the treatment of PD, such as a renewed interest in stereotactical surgery[39,40] or physiotherapeutical interventions[41], have also increased the need for reliable instruments for evaluating changes in motor function as a result of disease progression, or as a result of therapeutic interventions. To develop such instruments requires a sound understanding of the underlying pathophysiology of motor disturbances in PD, empirical data (clinical observations, or data from movement studies in PD), and/or a theory about coordination of movement which encompasses both normal and disordered movement.

Motor problems in PD predominantly concern the voluntary initiation and involuntary execution of complex movements[42]. The slower movements in PD can partly be attributed to what has been called the energising problem; a deficit in the ability to swiftly activate (and deactivate) a muscle[43,44]. However, the simultaneous and/or sequential execution of movements is more problematic than what may be expected based upon the slower execution of single movements[45,46]. Apart from these deficits, postural responses are less effective in PD[47,48], and anticipatory postural activity may be diminished or absent[49,50]. Thus, Parkinsonian motor problems increase especially when the complexity of a movement pattern increases, when movement execution requires fast movements, and when own movement activity or external perturbations disturb posture or equilibrium. Observations of disturbances in certain aspects of movement behaviour in PD patients (e.g. turning in bed, sitting down or rising from a chair, turning while standing or walking) suggest that a separate group of motor disabilities can be discerned in which Levodopa-therapy is only marginally effective[51,52]. Studies of basal ganglia function[53] and psychomotor tasks[54,55] suggest that the loss of motor function in PD must be particularly attributed to internal control mechanisms which fail to provide cues needed for the automatic execution of motor tasks. Clinical observations, and a limited number of studies, suggest that a conscious control of movement execution[41,56,57] and the presence of relevant sensory information[58-60], for example visual or auditive cues, may help Parkinson patients to overcome their problems in initiating and executing movements. However, cognitive activity which distracts from a conscious control of movement execution may markedly aggravate motor problems in PD. In regard to the gait of PD patients, a large number of deviations have been reported, e.g. a reduced walking speed, stride length and stride frequency[61,62]; an abnormally flexed posture; rigid trunk movements coinciding with a decreased arm swing[62]; abnormal postural activity during the initiation of gait[63]; and an abnormal foot strike[59,64].

Clinical analysis of human gait and movement

With recent advances in measurement technologies, and improved possibilities of automated data acquisition and analysis, the expectations concerning the contribution of movement analyses to clinical decision making have obviously increased. For rehabilitation purposes often the need exists for the assessment of motor function on a routine base, in order to determine disease progression, or the effects of therapeutic interventions. For the overall effect of therapy on daily life functions and quality of life, the assessment of motor function requires focusing the analysis on a disability level. However, to understand mechanisms causing disablement, several levels in the disablement process, i.e. links between pathology, impairment, functional limitations and disability, need to be analysed[65,66]. For understanding the effects of therapy or disease progression, clinical analysis of human gait and movement must be directed at the analysis of both impairments and functional limitations. Thus, both a routine based application of movement and gait analysis, and the use of these techniques in order to solve particular clinical research questions are necessary.

Routine based analysis techniques should meet a number of requirements which do not usually have to be met in a (clinical) research context. Apart from methodological criteria which any test procedure must fulfil (i.e. criteria regarding reliability, validity, responsiveness, and, for particular purposes, sensitivity and specificity) other criteria determine the usefulness in clinical practice. In order to satisfy criteria for clinical usefulness in daily clinical routine, the analysis techniques must perhaps meet practical requirements. More importantly, they must generate information which is relevant in clinical decision making. Regarding the relevance of biomechanical or any other tests to clinical decision making, it has been suggested[67] that test procedures must accomplish at least one of the following: 1) a differentiation between one of several possible disorders explaining some array of symptoms and signs (diagnosis); 2) a selection of the best among several treatment options; 3) a prediction of outcome (prognosis); 4) a determination of the severity of disease or injury (assessment); 5) an identification of patients at high risk for developing some serious problem (screening). In order to accommodate one or more of these demands, measured parameter(s) must correlate well with the patient's functional capacity, they must clearly distinguish between normal and abnormal, and they must not be directly observable and quantifiable. Moreover, measurements must not significantly alter the performance of the evaluated activity, and they must be accurate, reproducible, and cost-effective[67,68]. Unfortunately, the extent to which current biomechanical measures, or other parameters, meet any of these criteria can be questioned[67]. It appears that, with few exceptions[69], clinical studies of gait and movement do not yet substantially contribute to clinical decision making.

Disturbances of gait and balance in elderly

The ability to safely and independently walk is an important determinant of mobility. Any deterioration of gait function will lead to a disturbing decrease in mobility and often to a reduction of independent functioning in daily life. Moreover, gait deterioration and an increased risk of falling may lead to severe injuries, particularly in the elderly. Demographic changes leading to an increased elderly population have motivated a large number of studies of mobility, and gait and balance function in the elderly. These studies have shown that, although ageing per se may lead to changes at many levels of the neuromusculoskeletal system[70], gait and balance function are more strongly affected by pathology than by chronological age[71,72] (for reviews see Refs. 73-75). Ageing-related pathologies may lead to impairments at all levels of the neuromusculoskeletal system, and impairments at different levels may simultaneously contribute to gait problems in the elderly. Thus, in order to understand gait and balance problems in elderly, the impairments caused by different pathologies and their specific effects on gait and balance should be identified. Although impairments at a lower-level (i.e. musculo-skeletal problems, peripheral nerve lesions, or peripheral sensory problems) can be identified, this is often impossible with the higher-level impairments which result in gait and balance problems of a neurologic nature[76]. In addition to problems in specifying impairments, the effect an impairment at a certain level has on different aspects of gait often cannot be predicted. The latter fact is particularly due to the fact that even the principles of normal coordination of gait are only partly understood. These factors explain why the aetiology of gait and balance disturbances in individuals may be hard to identify.

When considering studies of disturbances of gait and balance in elderly in regard to criteria which determine their clinical relevance[67], it appears that the majority of these studies are directed at an assessment of function in dependence of age or specific pathologies. A diagnosis of the underlying pathology, selection of treatment, predicting outcome, and screening for the presence of presently known risk factors for falling[77,78], can usually all be established without additional gait or balance analyses[73-75]. The assessment of gait disturbances may both be directed at the identification of specific impairments, and the quantification of functional limitations in different pathologies. The former studies often address specific research questions, whereas the latter studies are directed at describing gait function, for example, in order to evaluate the effects of therapeutic strategies. An effective assessment of gait function requires valid measures of the quality of gait. Three important parameters which relate to the quality of gait are speed, stride length and stride frequency. These basic spatio-temporal parameters of gait can relatively easy be measured without large costs. Consequently, numerous studies of gait report these and other spatio-temporal parameters of walking. Spatio-temporal characteristics of walking are influenced by inter-individual differences in age, sex, and anthropometric characteristics[79-82]. When the walking pattern of an individual is repeatedly tested under the same conditions, the spatio-temporal characteristics are very similar. However, a change in individual characteristics, such as walking with or without shoes, may lead to changes in walking patterns[80]. Furthermore, different environments (e.g. different walking surfaces) and experimental conditions (e.g. treadmill walking versus overground walking[83]) may influence walking.

Studies of gait disturbances often focus on a comparison of walking at a spontaneously chosen speed in groups of patients and representative healthy control subjects (for examples, see Refs. 61, 62, 84, and 85). Despite the fact that speed dependent changes in different aspects of walking have been well described in healthy subjects[86,87], this approach is also often encountered in clinical gait studies in which, for example, forces, moments, kinematics, and patterns of muscle activity are reported (for exceptions, see Refs. 88 and 89). These studies do not always report the basic spatio-temporal parameters of gait. A number of disadvantages of this approach are obvious. First, with this approach, it is uncertain whether specific changes in gait parameters of a patient must be attributed to an almost invariantly reduced walking speed or to specific impairments. Thus, deviations in gait parameters can only indirectly be interpreted. Secondly, when the basic spatio-temporal parameters of gait are not reported, a direct comparison with results reported by other researchers is impossible. Thirdly, when the effects of different influencing factors on spatio-temporal parameters of gait are not correctly accounted for or when these effects remain unrecognised, an interpretation of deviating values in patients and a comparison with results obtained in other studies is questionable. Fourth, a statistical approach in which mean values of gait patterns in patients are compared to those of healthy controls suggests the existence of "normal gait patterns". However, as only aspects of the coordination of normal walking are understood, the definition of "normal gait patterns" is highly problematic.

Adaptability of human stepping patterns

From the preceding section, it can be concluded that both a valid assessment of gait quality, and the identification of specific impairments, require more than measuring walking at a spontaneously chosen speed only. Normally, our walking patterns can be adapted within a range of speeds, stride frequencies, and stride lengths. Moreover, basic coordination patterns can be adjusted. For example in order to avoid stumbling over an obstacle. Gait disturbances not only express themselves in abnormal gait patterns during walking at a spontaneously chosen speed, but particularly in the absence or reduction of the ability to adapt the gait pattern to changes in task demands. This adaptability to a large extent determines the quality of gait, and specific changes in adaptability may reflect changes in coordination due to specific impairments. Thus, both for the assessment of gait quality and for the identification of specific impairments, the study of the adaptability of gait patterns may be appropriate. The characteristics of gait adaptation to internal and external task demands are determined by both voluntary and involuntary mechanisms. Thus, both mechanical factors, and central control, will be involved in changes in basic gait patterns. As the way in which characteristic speed dependent changes in walking of healthy subjects are determined by passive and/or active factors is only partly understood, adaptability of gait deserves attention both in healthy subjects and in patients.

It can be concluded that to better understand normal and deviating coordination of human stepping patterns, it is necessary to study the adaptability of these patterns. Apart from studying adaptation to changes in walking speed, stride length, and stride frequency, the way basic coordination patterns are adapted to changes in internal or external task demands should also be studied. The work presented in this thesis aims to study adaptability of human gait by exploring the characteristics of stepping patterns as a function of speed, stride frequency, and stride length in different conditions. Several studies were performed in children, healthy young adults, healthy elderly subjects, and patients with Parkinson's disease, in order to identify characteristic differences in stepping patterns. These studies concentrate on variant and invariant aspects of stepping patterns in different conditions, and factors which explain these aspects. Chapter 2 presents a study of voluntary and involuntary adaptation of walking to auditory and visual cues which result in temporal and spatial constraints on stepping movements. Chapter 3 analyses the adaptation of stepping patterns in PD patients and healthy age-matched subjects. This chapter contains two studies, one is directed at the effects of increases in speed during overground and treadmill walking, and the other at the adaptation of walking to similar conditions as presented in chapter 2. Chapter 4 contains a study of stepping patterns of healthy subjects in treadmill walking conditions in which the left and right leg are offered different speeds (i.e. split-belt walking). Chapter 5 analyses the adaptation of stepping movements during treadmill walking and split-belt walking in children from four to ten years old. Chapter 6 describes and models patterns of pelvic displacement in dependence of frequency and amplitude of stepping movements. Finally, chapter 7 contains a general discussion of these studies.

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Chapter 2: Voluntary and Involuntary Adaptation of Walking to Temporal and Spatial Constraints

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Abstract

The effects of voluntary adaptations of the walking pattern were studied in 10 healthy subjects. Subjects had to adapt their walking pattern to external cues which resulted in either temporal or spatial constraints on walking. The results show that, whereas the normal walking pattern in each subject was characterised by a stable linear relationship between step frequency and step length, the adaptation of walking to external cues resulted in two different modes of walking: frequency modulation with a constant amplitude of leg movements in rhythmic walking, and amplitude modulation with a constant frequency of leg movements in walking with visually guided step length. The relative timing of support, swing and double support phases was remarkably different between the two conditions; only with changes in step length could a change in the relative timing of these phases be observed. The different modes of walking could not be explained by the nature of the constraints imposed by the instruction to adapt the walking pattern to the external cues. It can be concluded that the way step frequency and step length are controlled involuntarily varies with the employed voluntary strategies. The different modes of walking suggest the existence of specific supraspinal influences on lower level pattern generating mechanisms. The involuntary adaptation of the relative timing within the stride cycle may result from the interaction between (low level) central processes and the dynamics of the movement. *Gait & Posture 1995; 3: 13-18.*

Introduction

In the rehabilitation of gait disturbances resulting from supraspinal lesions (for instance, Parkinson's disease or hemiplegia), it is often attempted to improve walking by adopting new strategies of exerting voluntary control. However, little is known about the effects of differences in voluntary strategies on the locomotor pattern in either healthy subjects or patients. Therefore, the aim of the present study was to determine the influence of different voluntary strategies on spatio-temporal parameters of walking in healthy subjects.

The central control of walking and running in man is thought to be achieved by supraspinal mechanisms exerting control on lower level central mechanisms[1]. Voluntary and involuntary adaptation of the stride cycle allow for a high degree of flexibility of locomotion so that in different circumstances goal-oriented locomotion can be maintained. Although voluntary adaptation is possible (see for example Ref. 2), increases in walking speed are normally reached by a more or less invariant increase in both frequency and amplitude of leg movements[3-5]. The intra-individual relationship between frequency and amplitude of leg movements is stable and based on their experimental results some authors suggested that the step length/frequency ratio remains constant over a range of walking speeds[6,7].

A general assumption in this study is that variations in voluntary control lead to an altered functioning of lower level central processes which in turn leads to variations in basic temporal and spatial characteristics of locomotion like walking speed, step frequency, step length and timing of support and swing phases.

Methods

Ten healthy males volunteered to participate in this study; the volunteers were divided in two age groups: 25 to 30 years old (n=5) and 55 to 60 years old (n=5).

Subjects were instructed to walk on a 6 m long metal walkway with two metal strips under the sole of each shoe. This allowed recording of heel strike and toe off in both legs. In addition, the time between crossing two infra-red beams (placed 4 m apart on the central part of the walkway) was recorded. Recordings were made with a Siemens-Elema Mingograf and after A/D conversion the data was transferred to a PDP-11 computer. Successive data processing was done with interactive software on a micro computer. For each gait measurement at least 15 stride cycles were collected for the calculation of means and standard deviations of stride cycle parameters.

Gait measurements were made in three conditions:

Normal Walking at different speeds. Subjects were instructed to walk at five different speeds: extra slow, slow, normal, fast, extra fast. Thus, the actual walking speed could be determined by the subjects themselves. The five different speeds were once instructed in a random order and once in an increasing order.

Rhythmic Walking. In this condition, all subjects were instructed to synchronise their walking pattern to the rhythmic sound of an electronic metronome. Six rhythms were presented in the following order: 65, 80, 90, 105, 115, 125 beats per minute. In all subjects these rhythms led to step rates which were about equal to the audible rhythms.

Walking with visual guide. First the normal step length (S_n) of walking at a self chosen normal speed was determined. Then subsequent measurements were made according to the following principles: white stripes were attached to the walkway at fixed distances and subjects were instructed to step over the stripes. Measurements were made with distances between stripes of: S_n - 20 / S_n - 10 / S_n / S_n + 10 / S_n + 20 cm. In some individuals only three or four measurements were made in stead of the five mentioned above.

In all conditions with external cues, subjects could practice before recordings were made. Subjects were free to adjust step length during rhythmic walking or step frequency during walking with visually guided step length.

An analysis of the mean spatio-temporal parameters was performed by means of a statistical software package (SPSS/PC^{TM}). For each subject and for both groups of subjects, linear regressions were calculated to express the relationship between step frequency (N), step length (S), walking speed (V) and phase durations expressed as a percentage of stride cycle duration. This was done for all conditions. For the range of step frequencies obtained during normal walking, walking speeds were predicted by means of the (individual) linear regression functions of walking speed on step frequency. This was done twice; once with the regression functions obtained for the conditions in which walking speed was instructed in a random order, and once with the regression functions for the conditions where the five speeds were instructed in an increasing order. The two predicted speeds were used to determine the stability of the relationship between S and N with increasing walking speed in normal conditions.

Results

Despite inter individual differences in the measured absolute values, all subjects showed similar relationships between the analysed temporal and spatial variables in all conditions. The differences between young and old subjects can be summarised as follows; in all conditions where the choice for step length was free, the older subjects walked with somewhat smaller steps. Minor differences in the timing of support and swing phases could be attributed to these smaller steps. Table 1 illustrates the similarity of the trends between the analysed parameters in the young and old subjects by means of the calculated linear regression functions for both groups. The following presentation of results will be restricted to new findings concerning the adaptation of the walking pattern to external cues.

Walking speed, step frequency, and step length

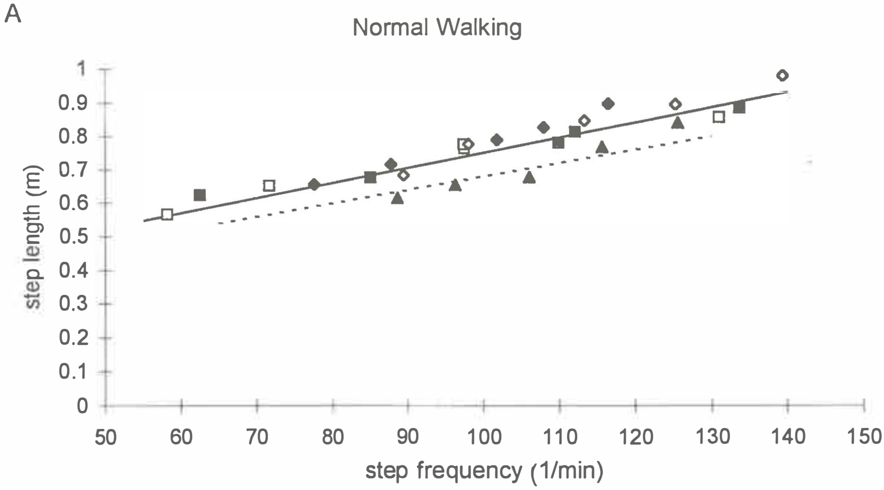
The way in which subjects were instructed to modulate their speed (random or increasing) hardly influenced the way in which step length and step frequency contributed to an increase in walking speed. The absence of changes in individual regression lines was confirmed by the following procedure; in case of equivalence of the results of the random instructions and the instructions given in an increasing order, the linear regression function of the predicted walking speeds (Vpred1 and Vpred2, see Methods) should be: Vpred1 = Vpred2. The actual regression function for the whole group of subjects was: Vpred1 = 0.998 * Vpred2 - 0.0028 ($R = 0.997$, $p < 0.0001$). Because walking speed is solely determined by step length and step frequency, this regression function implies that at self-chosen speeds the relationships between step frequency, walking speed, and step length are almost fixed.

In the following presentation of results, the data from the measurements in which walking speed was increased stepwise will be used as representing normal walking conditions. For convenience of interpretation, this data will be presented with either step frequency (N) or step length (S) as independent variables.

In figure 1, a comparison is made between normal walking and rhythmic walking. Figure 1b shows that the normal relationship between N and S is lost in rhythmic walking; here all subjects modulated N (according to the different rhythms offered) while maintaining S more or less constant (only in one of the five young subjects was there a slight but significant increase of S with N). This implies that an increase in walking velocity was reached by higher step frequencies only.

In figure 2b, it can be seen that this situation is reversed during walking with visual guide. With one exception, all subjects showed an almost constant step frequency with increasing step length (actually, step frequency was decreasing in most subjects; however, this decrease was never significant). In the one exceptional case where step frequency still showed an increase with step length, this increase was smaller than in normal conditions, and not significant.

The way in which modulations of step length and step frequency contribute to increases in walking speed can be expressed by means of length/frequency ratios. Figure 3 demonstrates the differences between the three conditions. When the conditions are compared within a similar speed range, remarkable differences can be seen; in normal walking the length/frequency ratio has somewhat higher values at the lowest walking speeds and tends to be constant at the three highest walking speeds, in rhythmic walking there is a clear decrease, and in walking with visual guide there is a clear increase in the length/frequency ratio.



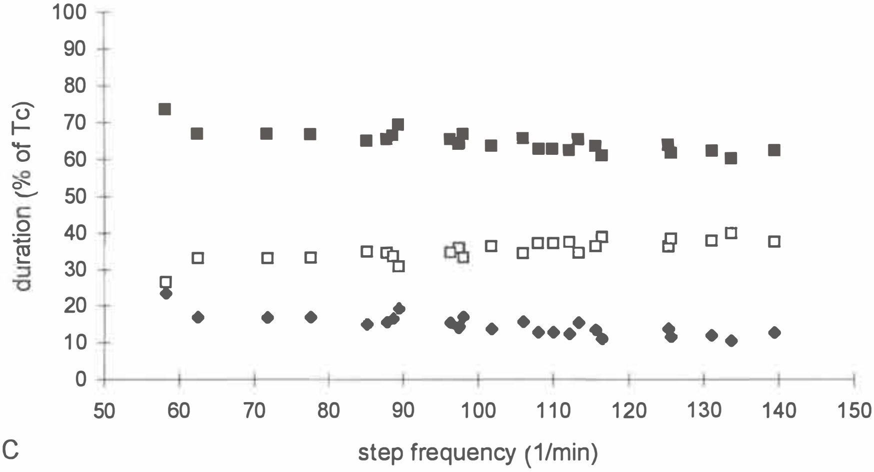
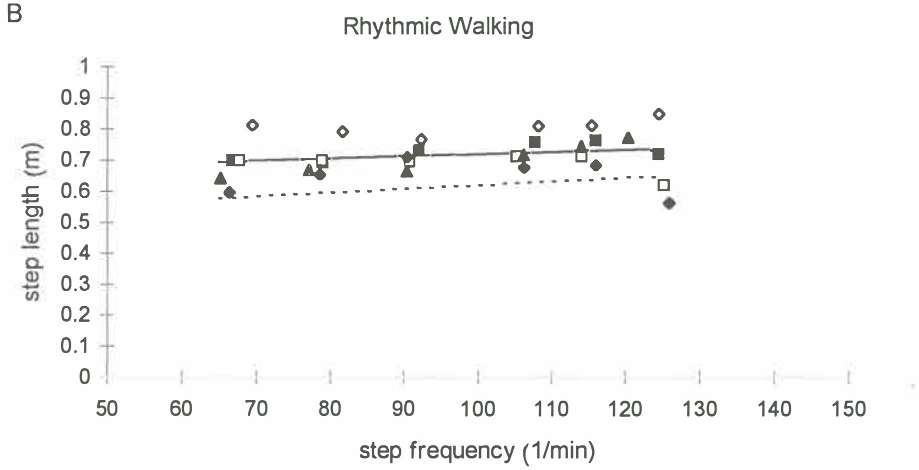
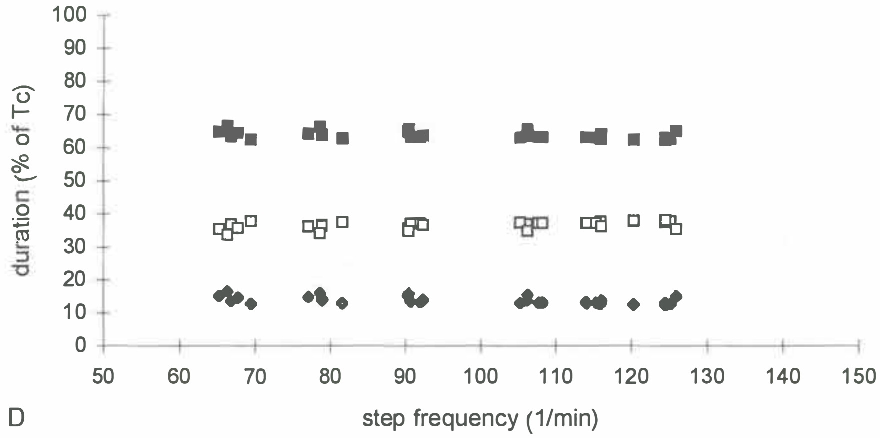


Figure 1: A comparison of rhythmic walking (right page) with normal walking (left page). In the two upper figures (1a and 1b), the different symbols indicate the relation between step length and step frequency in five young subjects. In the lower figures (1c and 1d), symbols represent the duration of support (■), swing (□) and double support (◆) phases expressed as a percentage of stride cycle duration (T_c). Bold lines (in figures 1a and 1b) indicate the linear regression functions for the young subjects (see also Table 1). Dashed lines represent the linear regression functions for the older subjects.


B
Rhythmic Walking
step length (m)
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
50
60
70
80
90
100
110
120
130
140
150
step frequency (1/min)


duration (% of Tc)
100
90
80
70
60
50
40
30
20
10
0
50
60
70
80
90
100
110
120
130
140
150
D
step frequency (1/min)

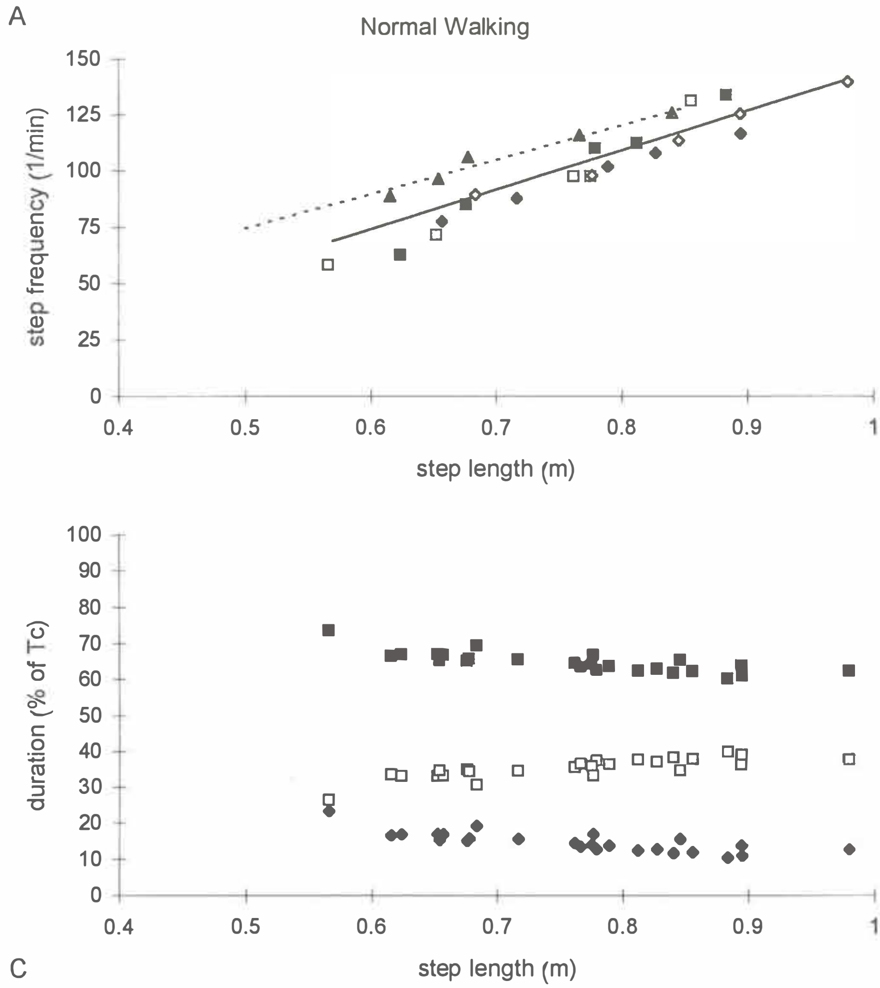
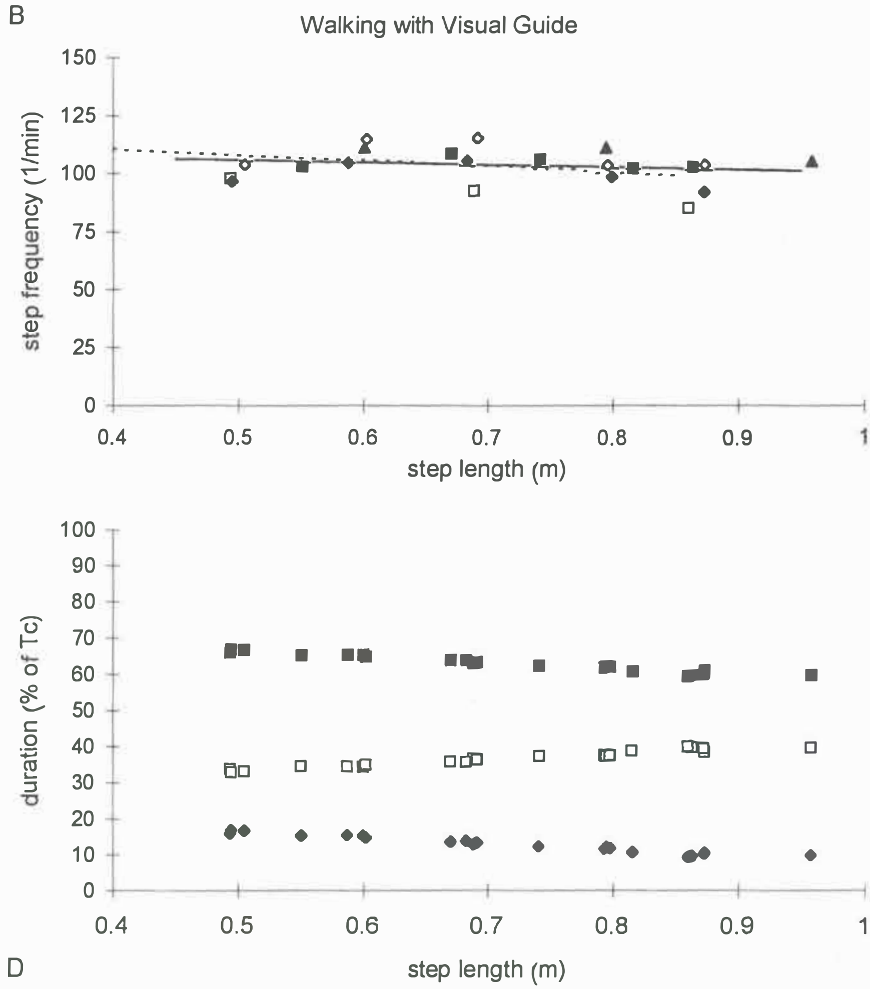


Figure 2: A comparison of walking with visual guide (right page) with normal walking (left page) Symbols and lines as in figure 1.


B
Walking with Visual Guide
step frequency (1/min)
150
125
100
75
50
25
0
0.4
0.5
0.6
0.7
0.8
0.9
1
step length (m)
duration (% of Tc)
100
90
80
70
60
50
40
30
20
10
0
0.4
0.5
0.6
0.7
0.8
0.9
1
D
step length (m)

	Normal Walking	Rhythmic Walking	Walking with Visual Guide
Young	**S = 0.0045*N + 0.30** *(R = 0.88, p < 0.001)*	**S = 0.00069*N + 0.65** *(R = 0.20, n.s.)*	
Old	**S = 0.004*N + 0.28** *(R = 0.78, p < 0.001)*	**S = 0.0012*N + 0.50** *(R = 0.31, n.s.)*	
Young	**N = 174.2*S - 30.3** *(R = 0.88, p < 0.001)*		**N = -10.3*S + 111.0** *(R = 0.20, n.s.)*
Old	**N = 153.5*S - 2.6** *(R = 0.78, p < 0.001)*		**N = -24.4*S + 120.5** *(R = 0.59, p < 0.01)*
Young	**T_{sup} = -0.11*N +75.8** *(R = 0.81, p < 0.001)*	**T_{sup} = -0.027*N + 66.2** *(R = 0.47, p < 0.05)*	
	T_{sup} = -20.3*S +80.2 *(R = 0.77, p < 0.001)*		**T_{sup} = -16.1*S + 74.6** *(R = 0.98, p < 0.001)*
Old	**T_{sup} = -0.10*N +75.7** *(R = 0.80, p < 0.001)*	**T_{sup} = -0.023*N + 67.5** *(R = 0.25, n.s.)*	
	T_{sup} = -24.3*S + 81.9 *(R = 0.96, p < 0.001)*		**T_{sup} = -12.0*S + 72.2** *(R = 0.73, p < 0.001)*

Table 1: Linear regression functions of the analysed temporal and spatial variables of both age groups. S is step length (m), N is step frequency (1/min) and T_{sup} represents the duration of the support phase expressed as a percentage of the stride cycle duration.

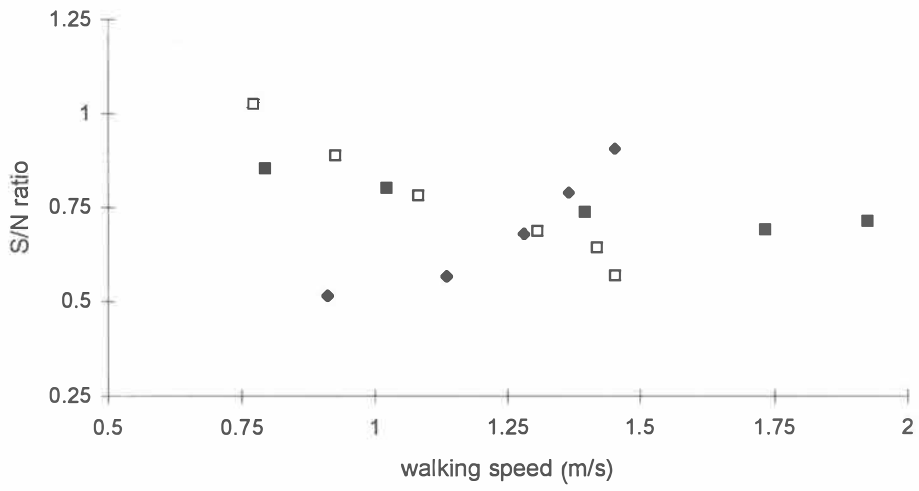


Figure 3: Mean ratios for step length/step frequency of five young subjects. Symbols indicate different conditions; normal walking (■), rhythmic walking (□) and walking with visual guide (◆).

Durations of support, swing, and double support phases

Although in all three conditions a decrease in support and swing phase duration was present with increasing walking speed the way in which the relative timing of phase durations (phase durations expressed as percentages of stride cycle duration) changed was remarkably different between the conditions (see Table 1 and figures 1 and 2). In figure 1d, it can be seen that contrary to normal walking conditions (figure 1c), the relative timing of stride-cycle events does not change with increasing step frequency during rhythmic walking. Again the situation during rhythmic walking contrasts strongly with walking with visual guide (figure 2d); the latter situation shows a similar decrease in support and double support percentages and increase of swing percentage as in normal walking (figure 2c).

Discussion

Our results on healthy subjects walking at different speeds are in agreement with earlier gait studies (see for example Refs. 3-5). From our results, it can be concluded that although inter individual differences exist, the intra individual relationship between step frequency and step length is stable under normal walking conditions. This stability has also been shown by other authors and may be related to the fact that at different walking speeds an optimal combination of step frequency and step length exists which results in minimal values for energy expenditure[8,9]. It can be concluded that only under normal walking conditions the step length/frequency ratio is similar to the constant ratio found by other authors[6]. However, this ratio is not perfectly constant but somewhat higher at low walking speeds and somewhat lower at high walking speeds (see figure 3). The remarkable changes in this ratio during rhythmic walking or walking with visual guide are surprising because in both conditions the choice for combining step frequency with step length (or vice versa) is free. If in the latter two conditions the length/frequency ratio would be determined by requirements for minimum energy consumption the same ratios as in normal walking would be expected. Thus, it can be concluded that the different voluntary strategies employed in both conditions (i.e. adapting walking to either temporal or spatial constraints) lead to two different modes of walking: frequency modulation (FM) in rhythmic walking and amplitude modulation (AM) in walking with visual guide.

Experimental evidence has suggested that central pattern generators (CPGs) are responsible for the generation of the sequences of muscle activations which underlie automatic stepping movements in animals and humans (for review see Ref. 1). It has been assumed that increases in walking speed are the result of a non specific increase in drive to these lower level CPGs[1]. This assumption is contradicted by the data presented in this paper. Our data rather suggest the existence of specific supraspinal influences on lower level mechanisms. Based on our data one can hypothesise that FM and AM are based on different neural control mechanisms.

The voluntary adaptation of the walking pattern to speed instructions or external stimuli coincided with involuntary adaptations of the stride cycle. Differences in timing of stride cycle events could be observed during FM and AM. The relative timing only changes with changes in step length; although a large range of frequencies (and velocities) is covered no change in relative timing occurs during FM only.

In subjects who perform auditorily cued alternating stepping movements on the spot, there exists a parallel decrease in support and swing duration with decreasing cycle time[10]. This is interpreted by some authors[10,11] as supporting the contention that an invariant timing of stride cycle events can be regarded as a characteristic property of a generalised motor program (see Ref. 11). In this view, a change in phase relations is an expression of a switch to another type of gait generated by another motor program with different phase characteristics. However, we consider it unlikely that the changes in relative timing during AM must be explained by the implementation of different motor programs. It was shown[12] that in conditions where the dynamics of walking were manipulated by selective loading of the movement apparatus that subjects achieved the same walking speed with the same frequency and amplitude of leg movements. The relative timing of support and swing phases changed according to the changes in dynamic conditions in which walking was performed. This shows that the timing of stride cycle events does not necessarily have to be a property of the central processes responsible for the modulation of leg movements during locomotion. Our results and the above mentioned results[12] rather suggest that phase relations within the pattern of leg movements emerge from an interaction of central processes with a biomechanical context.

What factors could then play a role in the observed differences in relative timing during FM and AM? One important constraint during walking at different speeds is the duration of the swing phase. It was argued[1] that, with a constant amplitude of leg movements it would be advantageous to keep the time it takes to swing the leg forward constant as well. In this situation the force required to bring the leg forward remains constant. Only an increase in the force required to reverse the backward movement of the leg during the support phase to a forward movement during the swing phase is necessary. However this constraint can be discarded as being the cause for the observed differences in relative timing because during FM this situation would lead to an increase in the relative timing of the swing phase with increasing step frequency.

Another possible cause of the differences in relative timing during FM and AM could be the displacement of the body during the support phases of each leg. In earlier gait studies[2,7], it has been shown that in normal walking conditions the forward displacement of the body during double support tends to be constant. If true this would lead to the differences in relative timing found during AM and FM. Assuming a more or less constant walking speed during the period of double support, the displacement of the body during double support (S_{bip}) can be estimated by multiplying the duration of double support (T_{bip}) by walking speed (V). Because V equals stride length (S_c) divided by stride duration (T_c) this gives the following equation: $T_{bip}/T_c = S_{bip}/S_c$. If both S_{bip} and S_c are constant this would lead to the invariant timing observed during rhythmic walking. If S_{bip} is constant but S_c is changing, as is the case in normal walking or walking with visual cues, this would lead to changes in the relative timing. Although in our subjects the displacement of the body during double support was not exactly constant, this parameter showed only a small variation in all three conditions. Thus, a tendency to keep this parameter constant could contribute to the observed differences in relative timing.

In conclusion: The presented voluntary and involuntary adaptations of the walking pattern reveal some properties of the coordinative mechanisms during human walking. The employed voluntary strategies result in either FM or AM and suggest the existence of specific supraspinal influences on lower-level pattern-generating mechanisms. The involuntary adaptations of the stride cycle probably result from the interaction from lower-level mechanisms with changes in the dynamical movement context. The invariant timing during FM might emerge from the activity of a CPG generating stepping movements at different frequencies without the usual changes in amplitude of leg movements.

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Chapter 3: Voluntary and Involuntary Adaptation of Gait in Parkinson's Disease

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Abstract

Voluntary and involuntary adaptation of gait in Parkinson's disease (PD) were studied in two separate experiments. In the first experiment, effects of changes in voluntary control were studied by asking PD patients and age-matched healthy subjects to adapt their walking pattern to visual cues resulting in spatial constraints, and auditory cues resulting in temporal constraints on stepping movements. In the second experiment, the adaptation to increases in speed during overground and treadmill walking was studied. Most patients were able to adapt their walking patterns in accordance with instructions. Notwithstanding consistent differences in step length, the adaptation to different conditions under study was highly similar in PD patients and healthy subjects. Only during walking with visually guided step length were the observed adaptations in PD patients less consistent. Contrary to these dissimilarities, the involuntary adaptation of timing of support and swing phases within the stride cycle was very similar between groups. In all conditions, only with changes in step length could a change in relative timing be observed. Our findings show that voluntary adaptation of gait is possible in PD, and that basic involuntary coordination mechanisms are preserved. The observed disturbances in stride length regulation probably reflect an inability to perform fast movements in PD. *Gait & Posture (in Press).*

Introduction

Gait disturbances are among the most incapacitating symptoms of Parkinson's disease (PD). Typical features of gait disturbances in PD include a reduced stride length and lower walking speed[1,2]. Moreover, changes in kinematic[1,2], kinetic[3,4], and electromyographic[5] gait parameters have been reported. In addition to changes in the gait pattern itself, problems related to the initiation[6], continuation, or termination of gait may arise. PD patients may experience difficulties in movement execution when gait is part of a complex pattern of activity (for example, zipping a coat while walking to a car and paying attention to traffic). In later disease stages, loss of motor function is predominantly determined by disturbances of balance control[7]. The reduced walking speed in PD seems to result from a deficit in producing an appropriate stride length rather than stride frequency[8]. Nevertheless, PD patients do increase their speed of progression by increasing both stride frequency and stride length in a manner similar to that of healthy adults[8,9]. However, PD patients seem incapable of achieving high walking speeds, and, even when a lower walking speed is taken into account, stride length does not reach normal values in PD patients. As a result, comparisons between gait parameters of PD patients and those of healthy subjects always reflect the effects of differences in speed, stride length, and frequency. For some gait parameters (e.g. the timing of support and swing phases), this can be overcome by statistically correcting for these effects. However, for other parameters (e.g. kinematic patterns, patterns of muscle activity, or joint moments) the use of appropriate experimental protocols which control walking speed[10,11], stride length, or stride frequency may be required. Walking speed may be easily controlled by instructing the subject to walk at different speeds or by using a treadmill. Stride length and frequency can be voluntarily adapted to specific instructions or external cues[12]. However, these changes in conditions may induce different walking patterns, both in healthy subjects and patients. For example, it has been shown that spatio-temporal parameters of treadmill walking in healthy subjects are consistently different from those during overground walking[13], i.e. subjects make smaller steps when walking on a treadmill than during overground walking at the same speed. Furthermore, voluntary adaptation of step length or frequency to external cues also induces a number of involuntary changes in the walking patterns of healthy subjects[12].

In regard to motor problems in PD, there are many indications that conditions involving the use of exteroceptive information may facilitate movement[8,14-17]. However, clinical observations of the motor disturbances in PD suggest that the effects of exteroceptive information may also be inhibitive, e.g. some PD patients experience problems when walking through a door opening or when trying to avoid an obstacle by stepping over it. The increased sensitivity for exteroceptive information in PD is in accordance with studies of cognitive and motor disturbances which suggest that at least a part of the loss of motor function in PD must be attributed to internal control mechanisms which fail to provide cues needed for the automatic execution of motor tasks[17,18]. Apart from facilitative effects of external cues on movement execution, changes in the way voluntary control is exerted may also be beneficial in PD. Attempts to alleviate typical Parkinsonian disturbances in the automatic execution of gross motor skills, by learning patients alternative movement strategies which require conscious control, have led to encouraging results[19-21]. Thus, both the use of exteroceptive information in movement execution and new strategies of exerting voluntary control may be of use in the rehabilitation of PD patients. As only a limited number of studies were directed at an evaluation of the effects of external cues and voluntary adaptation on movement execution in PD, their potential contribution to treatment of movement and gait disturbances in PD remains unclear. In regard to problems in initiating and continuing gait, it is also possible that treadmill walking may facilitate gait in PD. Yet, considering the changed visual input due to a stationary surrounding during treadmill walking, the reverse is also possible. To our knowledge, no studies exist which compare overground and treadmill walking in PD patients. Knowledge of gait of PD patients in all aforementioned conditions is not only necessary for an analysis of gait parameters in dependence of speed, stride length, or stride frequency, but also to determine whether new strategies of exerting voluntary control, the use of external cues, or treadmill walking, may be of use in the rehabilitation of PD patients.

The present study evaluates the voluntary and involuntary adaptation of gait to external cues, speed instructions, and treadmill walking in PD patients. For this purpose, two separate studies were conducted. Study I investigates the adaptation to temporal and spatial constraints, while study II focuses on the adaptation to increases in speed during overground and treadmill walking. The following issues will be addressed: first, it will be evaluated whether PD patients are able to adapt their gait patterns according to the instructions; second, the voluntary and involuntary adaptation of walking in PD will be compared to the adaptation in healthy subjects in order to evaluate whether spatio-temporal parameters of walking are changed similarly in PD patients; third, it will be briefly discussed whether the results indicate possibilities for the treatment of gait disturbances in PD.

Methods

Study I: Adaptation to temporal and spatial constraints

Ten PD patients (8 male and 2 female) with ages ranging from 44 to 74 years old participated in the experiments. The severity of PD, as measured by the modified Hoehn & Yahr (H&Y) disease rating scale[22], varied from I½ to IV. The walking patterns of PD patients were compared to those of 5 young and 5 age-matched healthy subjects walking in similar conditions. The ages of these two groups of healthy subjects varied between 25-30 and 55-60 years old, respectively.

Subjects were asked to walk on a walkway under *three conditions*; walking at self-chosen speeds, rhythmic walking, and walking with visual guide. All healthy subjects participated in two measurement sessions, whereas gait in PD patients was studied in only one session. In healthy subjects, *walking at self-chosen speeds* was studied in a first session by means of instructing subjects to walk at five different speeds (extra slow, slow, normal, fast, and extra fast). In a second session, walking at a self-chosen speed, and walking in conditions where subjects had to adapt their walking pattern, were studied similarly in both PD patients and healthy subjects. Three measurements of the walking pattern at a self-chosen "normal" walking speed were made; at the start of the experimental session, and after measurements of rhythmic walking and walking with visual guide. For the analyses of *rhythmic walking*, each subject was instructed to synchronise his or her walking pattern to the audible rhythm generated by an electronic metronome. In the healthy subjects, six different frequencies were presented with frequencies ranging from 65 to 125 beats per minute. In PD patients, first a frequency was used that was equal to the step frequency of the preceding measurement of walking at a self-chosen speed. Thereafter, frequency was stepwise increased to a target value of 100 beats per minute. Patients who easily managed this target frequency were presented with frequencies higher than 100 beats per minute. Thus, this procedure led to different numbers of measurements in different patients. In conditions where subjects *walked with a visually guided step length*, the subjects were instructed to step over white stripes which were attached to the walkway at fixed interstripe distances. In healthy subjects, measurements were made at five different interstripe distances. In PD patients, an interstripe distance equal to the step length in the preceding measurement of walking at a self-chosen speed was used. Thereafter, attempts to increase step length by stepwise increasing the inter stripe distances were made. Thus, also in this condition, the number of measurements differed between patients. Under all conditions, subjects could practice before gait measurements were made. So that patients were not fatigued by the number of measurements; they were allowed a short break as needed. After a break, measurements were continued. In the attempt to reach a higher step frequency or step length, patients were not stressed to the limit of their walking capability. When it was observed that a patient was not able to walk according to the instructions for a particular condition, gait measurements for that condition were stopped and the next part of the measurement protocol was started.

Durations of foot contacts were measured on a 6 m long metal walkway. Additionally, the time needed to walk the distance between two infra-red sensors placed 4 m apart was measured. A description of the used methods can be found in[12]. The recorded signals were used for the calculation of spatio-temporal parameters of walking. For each recorded stride cycle, the following temporal parameters were calculated: left and right step duration and the duration of the stride cycle, durations of the support and swing phases of each leg, and durations of the two double support phases. For each gait measurement, mean walking speed was calculated. Mean step length was calculated as the product of mean walking speed and mean step duration. For each condition, at least 15 stride cycles were used for the calculation of individual means and standard deviations of the spatio-temporal parameters.

Study II: Adaptation to increases in speed during overground and treadmill walking

Ten male PD patients participated in the experiments. Their ages ranged from 48 to 73 years old (mean age 61 years). Their leg lengths (measured as the distance from the top of trochanter major femoris to the floor) ranged from 0.86 to 0.98 m (mean 0.91 m). The severity of PD varied from I½ to III on the modified Hoehn & Yahr (H&Y) disease rating scale[22]. PD patients were compared to 8 age-matched healthy subjects with ages ranging from 44 to 73 years old (mean age 61 years), and leg lengths ranging from 0.85 to 1.01 m (mean 0.94 m).

Subjects were instructed to walk at three different speeds on a 10 m long walkway (overground walking), and at three matched and three fixed speeds on a treadmill. First, walking at a self-chosen normal speed was measured and, thereafter, repeated measurements of overground walking were made in which the subject was instructed to walk slowly, normally, and fast. During treadmill walking, three measurements were made at speeds which were individually matched to the measured speeds during overground walking at slow, normal, and fast speeds, respectively. Subsequently, measurements were made during walking at fixed speeds: 0.7, 1.0, and 1.4 m/s. Each subject could shortly practice treadmill walking, before measurements were made.

For measuring the durations of subsequent left and right foot contacts, each subject had to wear appropriately sized, light gymnastic shoes in which custom made foot switches were placed. Each foot switch consisted of two wired flexible metal layers separated by a non-conductive layer in which spaces were left out under the heel and forefoot. Thus, with appropriate pressure, the metal layers touched, and indicated foot contact. The use of these foot switches allowed identical detection of foot contacts during overground and treadmill walking. Measurements of overground walking were started, and subsequently stopped, by crossing the beams of two infra-red sensors placed 7 m apart at the central part of the walkway. For each speed condition, approximately five gait measurements were made. During treadmill walking, measurements were started, and stopped by the experimenter. For each speed condition on the treadmill, two measurements were made, each lasting approximately 20 seconds. Under all conditions, foot contacts were sampled at a frequency of 100 Hz. After A/D conversion, the recorded data were transferred to a micro computer for further analysis using custom made software. Identically to study I, spatio-temporal parameters of walking were calculated. For the calculation of an individual mean stepping pattern in a certain speed condition during overground walking, gait measurements were selected in which the mean speed did not differ more than 0.05 m/s. For each condition, at least five stride cycles were used for the calculation of individual means and standard deviations of spatio-temporal parameters. Mean stride length was calculated as the product of mean walking speed and mean stride duration.

In both studies, a statistical analysis of spatio-temporal parameters in different conditions was performed using a statistical software package (SPSS/PC™). Paired or group *t*-tests were used to compare within and between group effects. Linear regression functions were calculated to describe the relations between spatio-temporal variables in individual or groups of subjects.

Results

Study I: Adaptation to temporal and spatial constraints

General remarks

Due to balance problems, three patients (1 male and 2 female; all with a H&Y score of IV) could not fully complete the experimental protocol. Therefore, only the results of seven patients will be presented fully. These patients all were male, the severity of PD varied from H&Y I½ to III, and their ages ranged from 44 to 74 years old (mean age 55 years). Except the three PD patients with a H&Y score of IV, all subjects were able to adapt their walking pattern according to the conditions with external cues. During rhythmic walking all subjects synchronised their walking pattern to the presented rhythms. This was always achieved by letting the instant of left and right foot contact (i.e. heel strike in most healthy subjects) coincide with the pulse generated by the metronome. Thus, rhythmic walking led to step rates which, approximately, were equal to the frequencies generated by the metronome. During walking with visual guide, the foot was placed directly after the stripes attached to the walkway. Healthy subjects did not experience any difficulties in performing this task, even when step lengths of 0.9 m were required. However, some PD patients did have some difficulties. This particularly concerned the three patients already mentioned, but, at large step lengths, other PD patients experienced difficulties as well. In the presentation of results, only data are used of conditions in which PD patients were able to adapt their walking pattern conform instructions. As will be seen in the next sections, the voluntary adaptation of walking in accordance to the instructions given in different conditions, led not only to changes induced by the nature of the instructions; but also to changes in gait parameters other than the ones under voluntary control. Due to the nature of the conditions, the following presentation of results will report step length and frequency rather than stride length and frequency. Step frequency will not be reported in SI units, but in steps per minute.

Walking at self-chosen speeds

Figure 1 shows the relation between step length and step frequency during walking at self-chosen speeds. The regression lines clearly show that despite a similar increase of step length with step frequency, the age-matched-subjects walked with systematically lower step lengths than the younger subjects. Also, the range of step lengths and frequencies (and, thus, walking speeds) is somewhat reduced in the age-matched subjects. The individual values of PD patients show that step lengths are decreased in 9 out of 10 PD patients. Step frequencies are decreased in 7 out of 10 PD patients. Thus, in comparison to the healthy subjects, the walking patterns of the PD patients were characterised by a diminished step length (0.44 m for PD patients versus 0.69 m for the age-matched subjects, and 0.77 m for the young subjects) and step frequency (93.0 steps per minute versus 106.0, and 102.7 steps per minute, respectively). Consequently, walking speed in PD patients was low when compared to that of the healthy adults (0.70 m/s versus 1.22, and 1.31 m/s, respectively).

In all groups, minor changes in step frequency and length were observed in subsequent measurements of walking at a self-chosen "normal" speed. These changes were largest in PD patients, but even in this group the changes were rather small. After rhythmic walking, the step frequencies during the measurements of walking at a "normal" speed were somewhat higher, step length values remained the same or decreased, and walking speed did not change compared to the first measurement of walking at a "normal" speed. For PD patients, the mean values for these parameters were 96.9 steps per minute, 0.43 m, and 0.71 m/s, respectively.

When compared (paired *t*-tests) with the first measurement of walking at a self-chosen "normal" speed, these changes were significant for step frequency only ($p<0.01$). In all groups, the measurement of walking at a "normal" speed after walking with visual guide, did show a slight increase in step length. Step frequency remained the same in healthy subjects and was somewhat decreased in PD patients. Walking speed was slightly increased compared to that in the preceding measurement of walking in the same condition. For PD patients, the mean values for step frequency, length, and walking speed, during the third measurement of walking at a "normal" speed were 90.3 steps per minute (n.s.), 0.49 m ($p<0.05$), and 0.75 m/s ($p<0.05$), respectively.

The lower part of figure 1 indicates a similar decrease of support duration with increasing step frequency for the young and age-matched healthy subjects. However, the walking pattern of the young subjects is characterised by a somewhat shorter duration. With one exception, all PD patients have a longer duration of support than the duration of support as indicated by the regression function for the age-matched subjects. The patient with a shorter relative duration of support is the patient with close to normal values for step length and frequency (see upper part of figure 1).

Rhythmic walking

In figure 2, the changes in step length and relative timing of the stride cycle with increases in step frequency are shown. Seven PD patients were able to achieve rhythmic stepping rates higher than the target value of 100 steps per minute. In both groups of healthy subjects, the increase in step frequency hardly leads to an increase in step length. The slight increase in step length is not significant (see legends of figure 2). Again the older subjects have lower step lengths than the young subjects. Also in the individual PD patients step length almost remains constant over the individual range of step frequencies. Due to the approximately constant step length in each individual, the inter-subject differences in step length of PD patients remain constant over a range of step frequencies. Only two patients approximate normal values for step length and frequency, whereas the other patients have consistently lower step lengths at all step frequencies.

In the lower part of figure 2, the relative duration of the support phase is shown as a function of step frequency during rhythmic walking. Contrary to the decrease of relative support duration during walking at different speeds (figure 1), the relative duration of support is now almost constant while step frequency increases (see also legends of figure 2). Similar to walking at different speeds, the walking pattern of the young subjects is characterised by a somewhat shorter duration. Despite considerable differences between patients, the relative duration of support also remains approximately constant in the individual PD patients. Nine out of ten patients have longer durations of relative support in comparison to the age-matched subjects. The one exceptional case is one of the two patients with close to normal step lengths during rhythmic walking (see upper part of figure 2).

Walking with visual guide

Figure 3 represents the changes in step frequency and relative timing of the stride cycle with increases in step length during walking with a visually guided step length. Contrary to the positive relationship between step frequency and length during walking at different speeds, step frequency decreases with increasing step lengths during walking with visual guide. Seven PD patients were able to increase their step length. The mean maximal increase in step length relative to the step length at a self-chosen normal speed was +0.19 m (range 0.11 to 0.36 m).

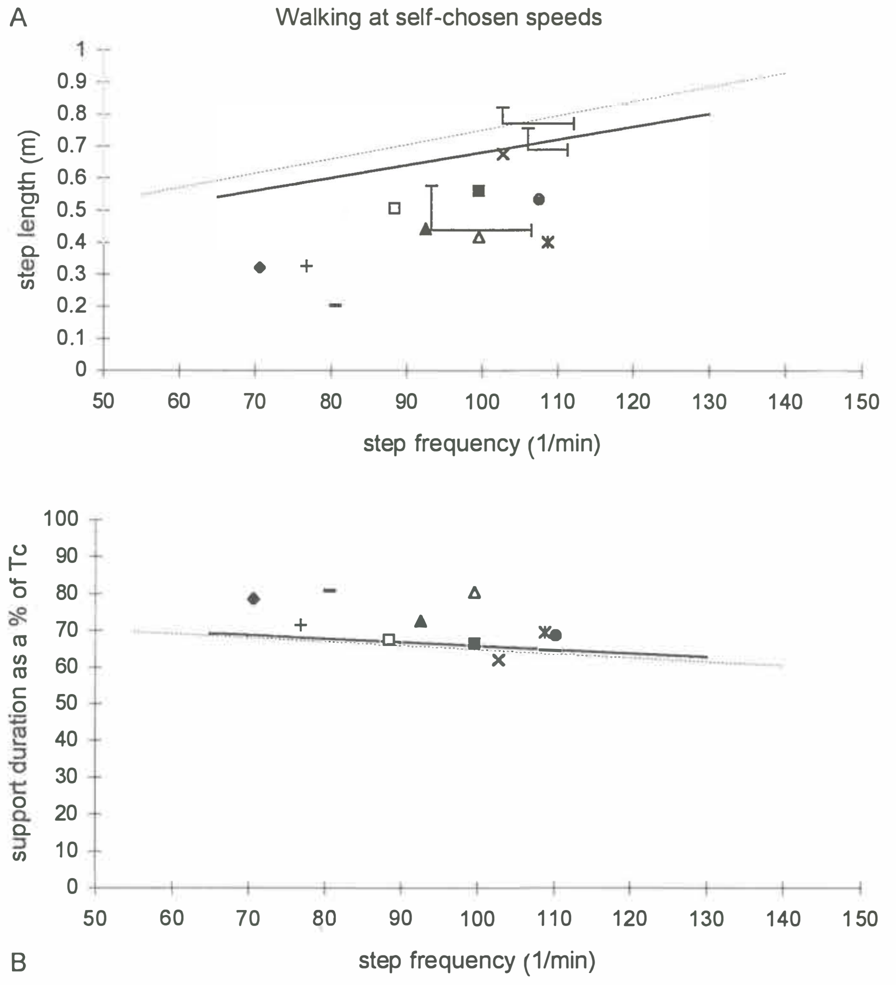

Figure 1a: Relation between step length (Sl) and step frequency (Sf) during walking at self-chosen speeds. The bold line represents the calculated regression function for the age-matched subjects (Sl = 0.0040 * Sf + 0.28 (R=0.78, p<0.001)), the dashed line represents the young subjects (Sl = 0.0045 * Sf + 0.30 (R=0.88, p<0.001)). Means and standard deviations (indicated by horizontal and vertical bars) of step length and frequency during walking at a self-chosen "normal" walking speed are indicated for both groups of healthy subjects, and the group of PD patients. Other symbols represent the individual values for step length and frequency in PD patients during the first measurement of walking at a self-chosen speed.

Figure 1b: Changes in the relative duration of the support phase (Tsup%) during walking at different speeds. Tc indicates stride duration. Lines and symbols as in Figure 1a. For the age-matched subjects: Tsup% = -0.10 * Sf + 75.7 (R=0.80, p<0.001), for the young subjects: Tsup% = -0.11 * Sf + 75.8 (R=0.81, p<0.001).

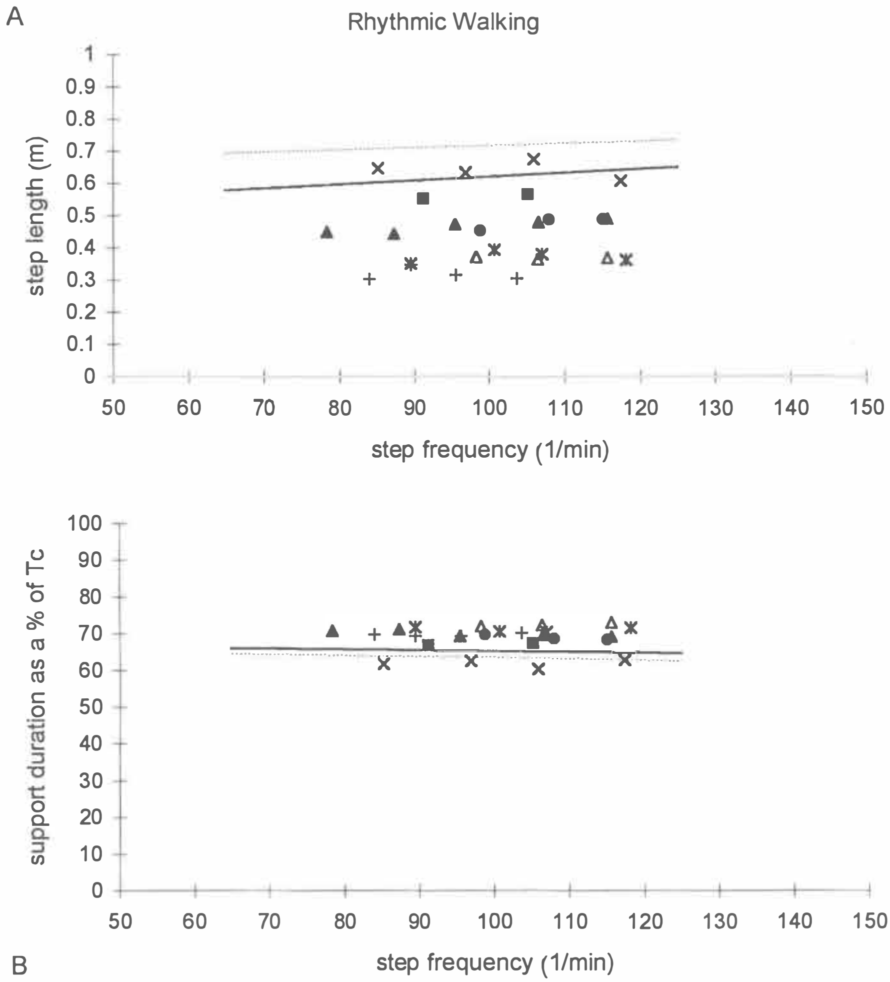

Figure 2a: Changes in step length (Sl) with increases in step frequency (Sf) during rhythmic walking. As in figure 1, both lines represent linear regression functions of the results in healthy subjects. For age-matched subjects the relation between step length and frequency is described by Sl = 0.0012 * Sf + 0.50 (R=0.31, n.s.), for young subjects Sl = 0.00069 * Sf + 0.65 (R=0.20, n.s.). The same symbols as in figure 1 are used to indicate specific PD patients.

Figure 2b: Changes in the relative duration of the support phase (Tsup%) with increases in step frequency (Sf) during rhythmic walking. Lines, symbols, and abbreviations as in Figure 1. For age-matched subjects: Tsup% = -0.023 * Sf + 67.5 (R=0.25, n.s.), for young subjects: Tsup% = -0.027 * Sf + 66.2 (R=0.47, $p<0.05$).

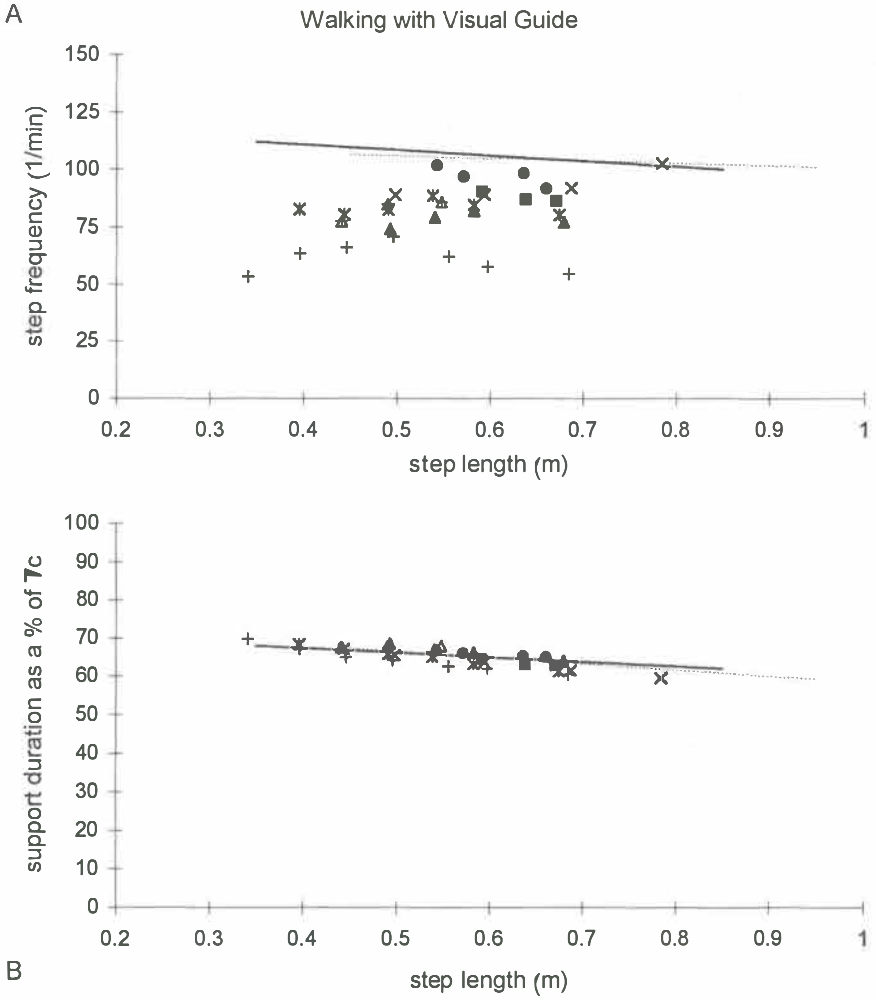

Figure 3a: Changes in step frequency with increases in step length during walking with a visually guided step length. As in figures 1 and 2, both lines represent linear regression functions of the results in healthy subjects. For age-matched subjects the relation between step frequency and length is described by Sf = -24.4 * Sl + 120.5 (R=0.59, p<0.01), for young subjects Sf = -10.3 * Sl + 111.0 (R=0.20, n.s.). Again the different symbols indicate the results in different PD patients (as in figure 1 and 2).

Figure 3b: Changes in the relative duration of the support phase (Tsup%) with increases in step length (Sl) during walking with visually guided step length. Lines, symbols, and abbreviations as in Figure 1. For age-matched subjects: Tsup% = -12.0 * Sl + 72.2 (R=0.73, p<0.001), for young subjects: Tsup% = -16.1 * Sl + 74.6 (R=0.98, p<0.001).

		Parkinson's Disease					Age-matched subjects				
		Overground		Treadmill			Overground		Treadmill		
Slow	V (m/s)	**0.47**	**0.20**	*0.49*	*0.20*	*n.s.*	**0.95**	**0.17**	*0.94*	*0.17*	*n.s.*
	1/Tc (Hz)	**0.63**	**0.11**	*0.90*	*0.14*	***	**0.77**	**0.05**	*0.87*	*0.05*	**
	Sc (m)	**0.74**	**0.27**	*0.56*	*0.23*	**	**1.23**	**0.17**	*1.08*	*0.18*	***
	Tsup (%)	**64.0**	**3.10**	*66.0*	*5.23*	*n.s.*	**60.4**	**0.99**	*60.4*	*1.45*	*n.s.*
Normal	V (m/s)	**0.83**	**0.27**	*0.82*	*0.27*	*n.s.*	**1.28**	**0.11**	*1.28*	*0.12*	*n.s.*
	1/Tc (Hz)	**0.80**	**0.10**	*1.02*	*0.16*	**	**0.90**	**0.04**	*0.96*	*0.03*	**
	Sc (m)	**1.02**	**0.28**	*0.83*	*0.29*	**	**1.43**	**0.12**	*1.34*	*0.13*	**
	Tsup (%)	**61.0**	**2.59**	*62.0*	*4.22*	*n.s.*	**59.2**	**0.69**	*59.3*	*1.62*	*n.s.*
Fast	V (m/s)	**1.29**	**0.28**	*1.25*	*0.23*	*n.s.*	**1.65**	**0.19**	-		-
	1/Tc (Hz)	**0.98**	**0.11**	*1.12*	*0.16*	*	**1.02**	**0.08**	-		-
	Sc (m)	**1.32**	**0.22**	*1.13*	*0.26*	*	**1.62**	**0.15**	-		-
	Tsup (%)	**59.6**	**2.47**	*58.3*	*4.74*	*n.s.*	**57.2**	**1.32**	-		-

Table 1: Mean spatio-temporal parameters of PD patients (n=10) and age-matched control subjects (n=8) during overground walking and treadmill walking at matched speeds. V indicates walking speed, 1/Tc is stride frequency, Sc is stride length, and Tsup indicates relative duration of support. Bold figures represent mean and standard deviations for overground walking, italics similarly represent treadmill walking at matched speeds. For both groups, significant differences between overground walking and treadmill walking at matched speeds are indicated by * ($p<0.05$), ** ($p<0.01$), or *** ($p<0.001$) (paired *t*-tests).

	Parkinson's Disease	Age-matched subjects
Overground walking	Sc = 0.76*V + 0.37 (R2 = 0.86)	Sc = 0.74*V + 0.51 (R2 = 0.86)
Treadmill walking at matched speeds	Sc = 0.84*V + 0.13 (R2 = 0.87)	Sc = 0.87*V + 0.25 (R2 = 0.91)
Treadmill walking at fixed speeds	Sc = 0.70*V + 0.33 (R2 = 0.67)	Sc = 0.69*V + 0.47 (R2 = 0.84)

Table 2: Linear regression functions for the relations between stride length (Sc) and speed (V) in the different conditions. For age-matched subjects, the regression functions for overground walking and treadmill walking at matched speeds are calculated for the data of walking at a slow and a normal speed only (see text).

In all individual healthy subjects, the increase in step length during walking with a visual guide led to a constant (or slightly) decreasing step frequency (not shown). This could not be observed in PD patients. In some PD patients, step frequency decreases after an initial increase. In other patients, a trend towards a continuously (slightly) increasing frequency can be seen. Two patients approximated normal values for step length and frequency, whereas others had consistently lower step frequencies at any particular step length.

In the lower part of figure 3, the relative duration of the support phase is shown as a function of step length during walking with visual guide. Age-matched and young subjects have a similar decrease of support duration. The decreasing relative support duration in the healthy subjects is closely approximated by the decrease in relative duration of support in PD patients. Differences between PD patients and healthy subjects are now minimal when compared with the observed differences during walking at different speeds and rhythmic walking.

Study II: Adaptation to increases in speed during overground and treadmill walking

General remarks

All age-matched subjects and PD patients were able to adapt their walking patterns according to instructions. All PD patients could perform treadmill walking at all matched and all fixed speeds. In some healthy subjects, the instruction to walk fast resulted in speeds which could not be matched by the treadmill. The following presentation of results will therefore not consider treadmill walking at a matched fast treadmill speed in the healthy subjects. Slight deviations between matched speed and speed during overground walking exist both in patients and healthy subjects (see Table 1). These differences are caused by the fact that each matched speed was the calculated mean of the gait measurements made for a particular speed condition during overground walking. However, not all measurements were used for the subsequent data analysis (see Methods). Therefore, small differences between matched treadmill speed and the calculated mean speed after data analysis may exist. The two measurements of each speed condition during treadmill walking did not yield statistically significant differences in stride duration in patients and healthy subjects. The second measurements were used for further calculations. The following presentation of results will report stride length and frequency data in SI units.

Overground walking at different speeds

Table 1 summarises the results of overground walking. In both groups of subjects, the different speed instructions lead to increases in walking speed, stride length and stride frequency, and a decreasing duration of support. Lower speeds and stride lengths in PD patients are apparent in all conditions. Also, stride frequency is lower in PD patients, but this difference with age-matched subjects decreases with increasing speed. The range in which PD patients modified their walking speed is somewhat larger than that of the age-matched subjects. This observation is also noted in the adaptation of stride frequency and stride length. Stride length actually increases more steeply in PD patients than in the age-matched subjects. Table 2 presents the linear regression functions which describe the relationships between stride length and walking speed for both groups of subjects. When comparing the functions describing overground walking in PD patients and age-matched subjects, the different coefficients (i.e. intercepts) illustrate the lower stride lengths at a particular speed (observe that the function for age-matched subjects describes walking at slow and normal speeds only). When individual data in PD are considered, a steeper increase in stride length in PD patients is evident. Values for the coefficients of linear regression functions describing the individual relationships between these parameters varied between 0.53 - 0.91 (slope), 0.1 - 0.61 (intercept), 0.976 - 0.999 (R^2) in PD, and between 0.43 - 0.72 (slope), 0.51 - 0.98 (intercept), 0.955- 0.999 (R^2) in age-matched control subjects. Mean values were 0.71 (slope), 0.43 (intercept), 0.988 (R^2) in PD, and between 0.55 (slope), 0.73 (intercept), 0.979 (R^2) in age-matched control subjects. These results clearly illustrate that all PD patients show linear increases of stride length with increasing speeds. Stride length is consistently lower but it increases more steeply with speed in PD. The individual linear regression functions represent the relationships between stride length and speed for individually different speed ranges. Hence, the comparison of overground walking between both groups, and between individual data, is made for different speed ranges and different starting speeds. Therefore, the comparison between walking patterns during the conditions with fixed treadmill speeds may yield other results (see next sections).

Overground versus treadmill walking

Table 1 compares overground and treadmill walking at matched speeds. In both groups, and in all speed conditions, treadmill walking induces walking with significantly higher stride frequencies, and lower stride lengths. However, this effect is somewhat stronger in PD. Thus, the typical differences between PD patients and age-matched subjects which were observed during overground walking become even more pronounced. During treadmill walking at a slow and normal speed, PD subjects have a slightly longer duration of support than during overground walking at the same speeds (however, these differences are not significant). These differences are not observed in the control subjects. The regression functions in Table 2 illustrate above mentioned changes in the relationships between walking speed and stride length. It can be seen from the coefficients of the linear functions that, in both groups, stride length during treadmill walking at matched speeds is lower than compared to overground walking. However, stride length increases more steeply during treadmill walking and, therefore, the differences in stride length between overground and treadmill walking become less with increasing speed. Nevertheless, in both groups stride length during treadmill walking remains smaller within the speed range investigated.

Age-matched subjects versus PD patients during treadmill walking at fixed speeds

The results in Table 3 show that within an identical speed range, PD subjects have smaller stride lengths and higher stride frequencies, but they adapt their stride length and frequency in an identical range as the healthy subjects. With respect to support duration, no significant differences can be seen between the two groups during treadmill walking. A remarkable fact is the change in coefficients of linear regression functions compared to treadmill walking at matched speeds in both groups of subjects (see Table 2). Although stride length is still smaller than during overground walking in both groups, the differences are drastically reduced. When individual data are considered, all PD patients show linear increases of stride length with increasing speeds. Coefficients of linear regression functions between these parameters varied between 0.33 - 0.88 (slope), 0.16 - 0.56 (intercept), 0.819 - 0.999 (R^2) in PD, and between 0.55 - 0.87 (slope), 0.26 - 0.68 (intercept), 0.941- 0.999 (R^2) in age-matched control subjects. Mean values were 0.70 (slope), 0.33 (intercept), 0.972 (R^2) in PD, and between 0.69 (slope), 0.47 (intercept), 0.987 (R^2) in age-matched control subjects. Therefore, it is obvious that within an identical speed range, the increase in stride length in PD patients is equivalent to that in age-matched controls.

		Parkinson's Disease		Age-matched subjects		
0.7 m/s	1/Tc (Hz)	0.89	0.14	0.74	0.09	*
	Sc (m)	0.81	0.13	0.96	0.11	*
	Tsup (%)	62.50	2.18	62.10	1.86	*n.s.*
1.0 m/s	1/Tc (Hz)	0.96	0.12	0.87	0.06	*
	Sc (m)	1.06	0.13	1.16	0.08	*
	Tsup (%)	60.30	1.83	60.40	1.10	*n.s.*
1.4 m/s	1/Tc (Hz)	1.10	0.18	0.97	0.05	*
	Sc (m)	1.30	0.18	1.45	0.08	*
	Tsup (%)	57.30	3.40	58.40	1.18	*n.s.*

Table 3: Means (and standard deviations) of spatio-temporal parameters of PD patients (n=10) and age-matched control subjects (n=8) during treadmill walking at fixed speeds. Abbreviations as in Table 1. Significant differences between both groups are indicated by * ($p<0.05$), ** ($p<0.01$), or *** ($p<0.001$) (grouped *t*-tests).

Discussion

The here presented studies were performed in order to evaluate voluntary and involuntary adaptation of gait to external cues, speed instructions, and treadmill walking in PD patients. The studies show that, despite their gait disturbances, most of the PD patients were able to adapt their walking pattern according to the instructions in different conditions. The only exceptions were PD patients with a H&Y rating larger than III; these patients were not able to perform rhythmic stepping movements, and during walking with visual guide they experienced overt balance problems when trying to increase their step lengths as indicated by the visual cues. Due to large differences in disease stages (H&Y ratings varied from I½ to IV), and due to slight differences in the experimental protocol (see Methods), our analysis of parkinsonian stepping patterns in conditions with external stimuli focused on a qualitative rather than statistic comparison of individual data of patients with stepping patterns of healthy subjects. Nevertheless, characteristic differences, and similarities, can be noted. In all conditions where the amplitude of leg movements was not imposed by the experimental protocol the PD patients had lower step lengths than the age-matched group. Likewise, the age-matched group had lower step lengths than the group of young subjects. In PD patients, step frequency during walking at a self-chosen speed often was lower than normal values. However, this decrease was less than the decrease in step length. In all healthy subjects, the different conditions with external cues lead to different modes of walking: frequency modulation (in combination with a constant amplitude of leg movements) in rhythmic walking, and amplitude modulation (in combination with a constant frequency) in walking with a visual guide (for a discussion of these remarkable changes in stepping patterns see Ref. 12). Whereas rhythmic walking induced walking patterns similar to those of healthy subjects, walking with visual guide did not induce a constant, or slightly decreasing, step frequency with increasing step length in PD patients. In the latter conditions, linear relations between frequency and amplitude were less frequently observed in patients. Although PD patients were able to substantially increase their step lengths, nearly all patients failed to do so with normal step frequencies. Thus, the patients made large but slow steps. However, the involuntary adaptation of support and swing phases was highly similar between groups; only with changes in step length could a change in the relative timing be observed. During rhythmic walking, the relative timing of support and swing phases remained constant as a consequence of the constant step length in this condition. The inter subject variability in PD patients during rhythmic walking was predominantly caused by between patient differences in step length. The near to normal duration of support in one of the PD patients can hence be fully attributed to his normal step length.

When the adaptation to different speeds during overground and treadmill walking is compared between patients and age-matched subjects, it can be concluded that in both conditions the walking patterns in PD patients are similarly adapted as in the age-matched subjects. Linearity of the relation between stride length and speed is preserved in PD. The only consistent difference is the lower stride length in PD patients. At any particular speed within the speed ranges studied, PD patients walk with a lower stride length and higher frequency. Our results confirm the known differences between treadmill and overground walking[13]. The treadmill induces similar changes in PD patients. Remarkably, the relation between stride length and speed has changed somewhat when walking with fixed speeds is compared to walking at matched speeds. In the latter conditions stride length is lower in both healthy subjects and in PD patients. A likely explanation for this fact is a habituation to treadmill walking during the period of the experiments. It has been reported that a habituation to treadmill walking takes place during the first minutes of treadmill walking, and that this habituation takes place more rapidly in subjects who are familiarised with treadmill walking in preceding sessions[23]. Although our subjects were shortly introduced to treadmill walking before measurements were made, this apparently was not enough to prevent habituation effects. The habituation in PD patients is almost identical to that in the healthy subjects. The similar adaptation of gait in PD patients and age-matched subjects during both overground and treadmill walking suggests no large facilitative or inhibitive effects of treadmill walking on gait in PD. In fact, during overground walking the PD patients were able to voluntarily increase their stride length more than during treadmill walking at matched speeds. Even the somewhat higher "fixed" treadmill speed of 1.4 m/s did not induce stride lengths which were larger than those during overground walking at a fast speed. Thus, it can be concluded that, in PD patients (with H&Y ratings $\leq$ III) treadmill walking does not induce other effects on spatio-temporal parameters of gait than in healthy subjects.

In previous studies of Parkinsonian gait, it was found that Parkinson patients have longer periods of double support[1,2,24]. Some authors have interpreted this result to be due to disturbances in balance control[25]. The present studies clearly show that PD patients do not have a longer duration of (bipedal) support. When controlled for reduced step lengths, the relative timing of leg movements seems normal in PD. Therefore, it can be questioned whether the observed longer duration of double support in PD is a consequence of a fundamental deficit in regulating stride length, or an adaptive phenomenon resulting from the need to compensate for problems related to balance control. The fact that in our study PD patients were able to increase their step lengths during walking with a visually guided step length confirm the hypothesis that a large part of the gait disturbances in PD patients must be attributed to a fundamental deficit in regulating stride length based on internal motor control mechanisms[8]. However, the three patients who were not able to complete the experimental protocol clearly were not able to do so because of their balance problems. As all other patients (all with H&Y ratings lower than IV) did not yet experience overt balance problems, our results do not allow for an unambiguous conclusion. L-Dopa medication leads to immediate effects on amplitude of leg movements during gait[26,27], but it does not seem to improve postural control[28,29]. This fact suggests that reduced step lengths in PD (and the inherently increased durations of bipedal support) are caused by problems in regulating stride length. These problems probably can be attributed to an inability to perform fast movements[30].

When our results are compared to recent studies in which long lasting improvements of gait as a result of training with rhythmic auditory stimulation[31], and visual cues[32] were reported, one striking difference is that, in our study, the immediate effects of walking in conditions with external cues upon subsequent measurements of walking at a self-chosen speed are rather small. Nevertheless, significant changes in specific gait parameters were present. After rhythmic walking, step frequency was increased, and after walking with visual guide, step length and walking speed were increased. Thus, it may be concluded from our results, and those of former authors, that a specific training is necessary to achieve longer lasting effects on specific gait parameters. The here reported changes in walking patterns resulting from the adaptation of walking to different external cues have not been reported by other authors. Differences in the used experimental protocols probably obscured these immediate effects in other studies. In regard to the mechanisms which play a role in the observed changes of walking in different conditions, we have argued that these results indicate the existence of specific supraspinal influences on lower level pattern generating mechanisms, and that (involuntary) changes in relative timing seem to result from the interaction between low level central processes and movement dynamics[12]. Our present results indicate that the latter mechanisms are intact in PD. However, supraspinal influences on lower level central mechanisms obviously are disturbed in PD, and the facilitative effect of external cues may lie in a compensation for defect internal mechanisms as suggested by several authors[8,17,18]. Nevertheless, the presence of external cues alone is insufficient to facilitate gait. A voluntary adaptation to external cues is required for gait to improve. This is also confirmed by the results of Morris et al.[32], they argue that the effects of visual cues on walking in PD appear to result from focusing attention on step length. Our results correspond with this interpretation; a voluntary adaptation, which required focusing attention on rhythmic stimuli, or visual cues, induced condition specific changes in walking both in PD patients and healthy subjects. The required attentional effort possibly also explains the involuntary changes in coordinative patterns in healthy adults; specifically, the attentional effort may interfere with normal control mechanisms at a supraspinal level and thus lead to spatio-temporal patterns of walking which deviate from the normally very stable, and energetically optimal, linear relationships between step length and frequency[12].

In conclusion: PD patients (with H&Y $\leq$ III) are able to adapt their walking patterns in accordance with speed, and stride length, or frequency instructions. Notwithstanding consistent differences in stride length, the adaptation to the conditions under study are highly similar in PD patients and healthy subjects. Thus, these conditions may be of use in studying the speed, amplitude, and frequency dependency of gait parameters in PD. With respect to possibilities for therapeutic interventions, a voluntary adaptation to external cues is required for effects on gait, and lasting relevant effects seem to require training. The immediate effects of voluntary adaptation of walking to external cues are changes in the relationship between spatio-temporal parameters of gait in healthy subjects, and similar condition specific effects in PD patients.

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Chapter 4: Adaptability of the Human Stride Cycle during Split-Belt Walking

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Abstract

The adaptation of leg movements to split-belt conditions was studied during treadmill walking in 10 healthy subjects. Four different belt speeds (0.5/1.0/1.5/2.0 m/s) were offered in all possible combinations for the left and right leg. All subjects automatically adapted to split-belt conditions. Although in all subjects and in all conditions the adaptation of leg movements within the stride cycle resulted in a walking pattern, the stepping pattern in conditions with extreme speed differences could be best described as limping. Regardless of the speed combination offered to the left and right leg, the adaptation to differences in left and right belt speed always involved: (1) Changes in stride cycle duration. The stride frequency in split-belt conditions was always intermediate to the values found during normal walking at speeds corresponding to the left and right belt speeds. There was a tendency towards the normal value of the faster belt, however. (2) An asymmetrical adaptation of the amplitude of leg movements. The support length of a leg generally tended towards the normal value for the speed offered to this leg. Thus, within the same cycle duration, the leg on the faster moving belt always made larger amplitudes than the slower leg. (3) An altered timing of support, swing and double support durations within the stride cycle. An increase in the relative duration of the support phase of the slower leg and the swing phase of the faster leg allowed for the larger amplitude of leg movements of the faster leg. Contrary to the asymmetric adaptation in support and swing durations the adaptation in the durations of both double support periods were more or less symmetric. The results of this split-belt walking study point to a high degree of flexibility in the relative timing of leg movements during walking. It is argued that the adaptation to split-belt conditions involves the tuning of low level coordinative mechanisms to the specific task requirements during split-belt walking by means of the afferent feedback resulting from the movement pattern. *Gait & Posture 1995; 3: 250-257.*

Introduction

In the last decades the control of locomotion and other bilateral rhythmic movements has become the subject of research programmes of different branches of science (for examples, see Refs. 1-4). Many of these research programmes focus on interlimb coordination and, although different theories and concepts may underlie studies of interlimb coordination, these research programmes have shown that similar principles of coordination can be recognised in movement patterns as different as locomotion, postural control and bimanual motor performance[4].

Studies directed at the neural control of locomotion led to the development of the central pattern generator (CPG) concept (for review see Ref. 5) and stressed the importance of different kinds of afferent information in the generation of coordinated movement patterns[1,4,5]. Experimental work on animal locomotion generated the hypothesis that each limb is controlled by its own CPG and that CPGs of different limbs interact via interneural networks[5,6]. In line with this hypothesis, it is assumed that the animal is able to perform several types of alternating and non-alternating gait or to walk along a curved path by means of different ways of coupling the actions of these separate locomotor generators. Non-linear changes of behaviour, like in this case the transition from an alternating to a non-alternating type of gait, have been of central interest in studies directed at the dynamics of pattern generation in complex systems. The latter "dynamical approach" has shown that non-linear changes in interlimb coordination can be induced by changes in control parameters (in rhythmic movement execution, the latter could for example be cycle frequency), that these changes are accompanied by an increased variability in movement execution, and that the phase difference between limb movements (the relative phase) is an important parameter in describing these non-linearities. This dynamical approach has stressed the importance of the concept of self-organisation in explaining rhythmical phenomena in biology (examples or reviews of this approach to interlimb coordination can be found elsewhere, see Refs. 3, 7 and 8).

Although human bipedalism differs from the quadrupedal gait patterns observed in animal locomotion, some basic characteristics of the human stride cycle are similar to animal locomotion[5,9,10]. These similarities suggest that the basic control mechanisms are the same for human and animal locomotion. However, contrary to quadrupedal gait patterns, human bipedalism requires a very strict coupling between the movements of both legs and although there is a high degree of flexibility in the control of frequency and amplitude of leg movements[11,12], it is unclear in how far human walking allows a flexible control of left and right leg movements. It has been shown that already in infants unable of unsupported locomotion, there exists a tendency to perform alternating leg movements when stepping patterns are evoked by a treadmill[13,14]. It can be questioned whether an alternating pattern of leg movements can be maintained when an extreme adaptation of the mature walking pattern is required such as when walking on a treadmill with split-belts, i.e., on a treadmill which offers different speeds to both legs.

In the present study, the split-belt walking paradigm is used for studying interlimb control during human locomotion. This paradigm is not a purely artificial way of putting the flexibility of interlimb control to the test; in daily life walking in a circle or turning around require the inner and outer leg to cross different distances in the same time, thus leading to speed differences for both legs. This article is directed at an analysis of the temporal and spatial adaptations of the human stride cycle in split-belt conditions. The interlimb coordination during split-belt walking will be described in terms of left right differences in spatio-temporal parameters like stride frequency, support length, and durations of support and swing phases. The stability of the coordinative patterns in different treadmill walking conditions will be evaluated by means of coefficients of variability. It will be analysed whether the adaptation to split-belt conditions involves the maintenance of an alternating mode of locomotion (this could be either walking or running), or that a non-alternating mode of locomotion (for instance, galloping[15]) is adopted as an adaptive strategy. The results of this study will be compared to the results of similar animal experimental work.

Methods

Ten healthy males volunteered to participate in this study. Their ages ranged from 21 to 29 years old (mean age 26.7, standard deviation (sd) 2.8 years). Their leg lengths (measured as the distance from the top of trochanter major femoris to the floor) ranged from 0.84 to 1.03 m (mean leg length 0.94, sd 0.05 m) and their body masses ranged from 57 to 91 kg (mean body mass 73.7, sd 10.8 kg).

By means of a split-belt treadmill[16], subjects were offered all possible left-right combinations of four different belt speeds: 0.5 / 1.0 / 1.5 and 2.0 m/s. This added up to 16 different walking conditions. In all conditions, the subject stood on the treadmill and the left and right belt speeds were gradually increased from zero until they reached their end velocities. During the gradual increase of both belt speeds the subject started walking. Subsequently, both belt speeds remained constant for 1 minute and thereafter both belt speeds were gradually decreased until the treadmill speeds were zero. Conditions were offered in the same order for all subjects. First, the conditions in which left and right belt speed were equal (normal treadmill walking) were offered in an increasing order. Thereafter, the split-belt conditions were offered. The latter was done in such a way that, starting with the lowest left belt speed, all higher right belt speeds were offered. Subsequently the left belt speed was increased and again all higher right belt speeds were offered. After all conditions with a higher right belt speed were offered, the same conditions were offered vice versa (in these conditions the left belt was always faster). Subjects were asked to look straight ahead during all treadmill walking conditions. However, no attempts were made to prevent subjects from looking at their feet and the treadmill belts. Before data collection took place, subjects were familiarised with both normal treadmill walking and walking in split-belt conditions. This was done by asking the subjects to walk in two normal treadmill walking conditions (0.5 and 1.0 m/s) and two split-belt conditions (left belt speed at 0.5 m/s and right belt speed at 1.0 and 1.5 m/s, respectively). Each of these trials also lasted little more than 1 minute. Some subjects were offered an extra trial, but the total time used for familiarising subjects with the different treadmill conditions never exceeded a period of 5 to 10 minutes. During data collection the subjects wore short trousers and light foot wear.

In each condition data recording took place during the 60 s in which both belt speeds were constant. By means of Kistler force plates which were placed under the two treadmill belts the force exerted on the treadmill could be measured for each leg. Force signals were amplified (floating-input amplifier; bandwidth 3-1000 Hz; sensitivity 2mV/V) and transferred on-line to a micro computer system via an A/D converter sampling at 500 Hz. A more detailed description of methods of data recording can be found elsewhere[16].

The force signals of the left and right treadmill belt were analysed by means of special purpose software. The onset of the force signal of the right belt (i.e. the impact of the right leg) was used as a trigger for selecting stride cycles. From the recorded force signals, it was possible to determine stride cycle duration (T_c) and the durations of left and right support (T_{sup}), swing (T_{swi}) and double support (T_{bip}) phases. In the support phase of a leg, the leg is used to support body weight whereas during the swing phase the leg is moved forward. Thus, the support phase could be measured as the period during which a leg exerted force on the treadmill belt and the swing phase could be measured as the absence of exerted force. Stride cycle duration was defined as the time between two successive impacts of the right leg. During a stride cycle two double support periods can be measured. One double support phase is characterised by the transfer of body weight from the left to the right leg (T_{bip-lr}) and the other by a weight transfer from the right to the left leg (T_{bip-rl}). The double support phases could thus be measured as the durations of simultaneous support on the left and right leg. From the last part of the recording periods, 20 successive stride cycles were selected for calculating means and standard deviations of the different stride cycle parameters. This was done for each subject and each condition.

For further calculations and a statistical analysis (group means, ranges and standard deviations, *t*-tests, MANOVA repeated measures and Scheffe tests) a statistical software package was used (SPSS/PC™). For all conditions and all individuals, a coefficient of variation was calculated to quantify the individual variability in stride cycle duration. This coefficient was calculated by expressing the standard deviation of the stride cycle duration in a particular condition as a percentage of the mean stride cycle duration in that condition. Amplitudes of leg movements were estimated by calculating the support lengths of each leg. The support length is an indication of the distance the leg is displaced by the moving belt, it can be estimated by multiplying the duration of the support phase of a leg with the belt speed offered to that leg. Phase lags were calculated to quantify changes in the timing of stride cycle events. This was done by expressing the duration of the time period between identical events in the left and right leg as a proportion of stride cycle duration. Thus, in case of a perfectly symmetrical alternating gait pattern, a phase lag of 0.5 should be acquired for all repetitive events in the stride cycle.

The possibility of differences in the adaptation of temporal parameters to "mirrored" conditions (that is: conditions with the same combination of speeds but either the left or the right leg being the leg on the faster belt) was statistically evaluated by means of *t*-tests pairs. This procedure involved comparing stride frequencies in all mirrored conditions and a comparison of the temporal parameters of the left leg with those of the right leg in similar functional conditions. The aim of this procedure was to evaluate a possible asymmetrical adaptation of stride cycle duration and left and right support and swing durations in mirrored conditions.

A two-factor repeated measures design with subjects as replications was used to evaluate whether the changes in stride cycle parameters as a result of changes in left and/or right belt speed were significant. All conditions were included in the analysis. Thus, each factor consisted of four levels (the four belt speeds). The effect of each factor and the interaction effect of both factors were determined. To indicate which levels contributed to significant main or interaction effects, post hoc comparisons were carried out by means of Scheffe tests (at a significance level of 0.05).

Results

Despite large differences in left and right belt speed, all subjects were able to automatically adapt their walking pattern. As could be expected in a locomotion task with the left and right treadmill surfaces moving at different speeds, some subjects used vision to control foot placement. However, all of our subjects managed to walk without the necessity of a continuous visual control of foot strike. In all subjects, a rhythmic alternation of left and right steps was reached within 10 to 20 s. Even a four fold speed difference (0.5 - 2.0 m/s) did not lead to problems for any of the subjects. The adaptation to split-belt conditions involved changes in all analysed spatio-temporal parameters of the stride cycle. Some subjects noted that they had to readapt to the normal walking situation after walking in split-belt conditions.

In the following presentation of the adaptation to split-belt conditions, the leg on the faster moving belt will be indicated as the “fast” leg and, vice versa, the leg on the slower belt will be indicated as the “slow” leg. Conditions with the same speed for the left and right leg will be indicated as normal walking.

Adaptations of the duration of the stride cycle

During normal walking an almost linear increase in stride frequency ($1/T_c$) with increasing belt speed was seen. In split-belt conditions, all subjects modified their stride frequency to a value intermediate to the stride frequencies found during normal walking at the corresponding speeds (see figure 1). The repeated measures analysis revealed significant main effects (for left belt speed: $F(3,27)=135.47$, $p<0.01$, for right belt speed: $F(3,27)=101.70$, $p<0.01$), and a significant interaction effect of belt speeds ($F(9,81)=6.28$, $p<0.01$). The stride frequency in a given split-belt condition tended more to the normal value for the fast speed. The adaptation of the stride cycle duration (T_c) did not significantly differ when for a certain speed combination the left or the right leg was on the faster moving belt. The latter was true for all speed combinations.

In normal walking conditions, the variability in stride cycle duration decreased with increasing belt speed. The mean coefficient of variation of all subjects significantly decreased from 2.4% at 0.5 m/s to 1.1% at 2.0 m/s. This effect on cycle variability was also seen in split-belt conditions; the combination of the higher belt speeds did lead to a lower variability than the combination of the lower belt speeds. The repeated measures analysis revealed that the decreasing variability as a result of increases in left and/or right belt speeds were significant (for left belt speed: $F(3,27)=8.30$, $p<0.01$ and for right belt speed: $F(3,27)=9.98$, $p<0.01$). Also the interaction of both belt speeds significantly led to a further decrease in variability ($F(9,81)=7.53$, $p<0.01$).

A post hoc analysis showed that during combinations of the lowest left belt speed (0.5 m/s) with higher right belt speeds the variability in stride cycle duration remained constant whereas the combination of a right belt speed of 0.5 m/s with higher left belt speeds did lead to a variability which was slightly higher than the variability during normal walking at the lowest speed (the mean coefficient of variation was 2.7% at left belt speeds of 1.0 and 1.5 m/s and 3.0% at a left belt speed of 2.0 m/s). However, the latter increases in variability were not significant and, as the effect of increases in right belt speed, the overall effect of increases in left belt speed was a decrease of the variability in stride cycle duration.

Adaptations in the amplitude of leg movements; support length versus belt speed

During normal walking conditions, support length increased linearly with increasing belt speed. In split-belt conditions, the adaptation of the stride cycle duration coincided with an asymmetric adaptation in the amplitude of movements of the "fast" and "slow" leg. This manifests itself in the left-right differences in the calculated support lengths (see figure 2). The support length of a leg tended to vary most with the belt speed offered to this leg. However, the smaller changes in right support length when left belt speed was varied were also significant. The repeated measures analysis revealed a strong main effect for right belt speed ($F(3,27)=532.40$, $p<0.01$), a small but significant effect for left belt speed ($F(3,27)=10.33$, $p<0.01$) and a significant interaction effect of both belt speeds upon right support length ($F(9,81)=19.57$, $p<0.01$). When left support lengths were compared with right support lengths in similar conditions, the existing small differences were never significant.

In figure 2, it can be seen that when a leg was offered a constant low belt speed (0.5 m/s) the support length of this "slow" leg decreased when contralateral belt speed was increased. For moderate speed differences, the support length of both legs tended to have normal values but the combination of a high speed (2.0 m/s) with lower contralateral belt speeds resulted in a decreasing support length of the "fast" leg. Thus, conditions with larger speed differences (combining 0.5 m/s or 2.0 m/s for one leg with other speeds) always resulted in a support length which was shorter than normal values in both the "fast" and the "slow" leg.

Relationships between belt speed and durations of support, swing and double support phases

In figure 3, a curvi-linear decrease of swing duration can be seen with increasing belt speed during normal walking conditions. During split-belt conditions, the swing duration varied mostly with contralateral belt speed and tended to be constant or slightly increase with ipsilateral speed. Again, the repeated measures analysis shows that the effects of changes in left and/or right belt speed on right swing duration are significant. The effect of left belt speed on the duration of the right swing phase is strongest ($F(3,27)=74.94$, $p<0.01$), but also the modest effects of right belt speed ($F(3,27)=33.34$, $p<0.01$), and the interaction of left and right belt speeds ($F(9,81)=5.59$, $p<0.01$) are significant. As with the duration of the stride cycle and the calculated support lengths, minor differences in the absolute durations of the support or swing phases of the "fast" leg when either the left or the right leg was the "fast" leg were never significant.

Whereas contralateral belt speed strongly influenced swing duration (the latter always decreased with increasing contralateral belt speed), support duration showed an inverse trend. Support duration decreased with ipsilateral belt speed and tended to vary less with contralateral belt speed (as the duration of the support phase of both legs is already represented in the calculated support lengths of figure 2, no additional figure of the changes in support duration is presented).

The above mentioned adjustments of support and swing durations lead to an altered relative timing in the stride cycle which allowed for different amplitudes of leg movements within the same stride cycle duration. An increase in the support duration of the "slow" leg coincided with a longer duration of the swing phase of the "fast" leg. Obviously, the relative durations of the swing phase of the "slow" leg and the support phase of the "fast" leg showed inverse trends. Thus, the "slow" leg provided support long enough to allow for the larger amplitude of leg movements in the swing phase of the "fast" leg. The adaptations in the support and swing durations of both legs forced the "slow" leg to cover its support length in a disproportionately shorter swing duration. This resulted in an asymmetry of leg movements which was speed dependent; the larger the speed difference the greater the asymmetry. In conditions with large speed differences this manifests itself as limping.

In figure 4, the adaptation in the two double support phases are shown. One double support phase can be characterised by the transfer of body weight from the left to the right leg (T_{bip-lr}) and the other by a transfer from the right to the left leg (T_{bip-rl}). The durations of the double support phases decreased curvi-linearly with increasing belt speed in normal conditions. Contrary to the adaptation of support and swing phases to split-belt conditions, both double support phases are adapted in a similar fashion. Increases in ipsilateral or contralateral belt speed have the same effect; both double support durations decrease with increasing speed. Changes in T_{bip-rl} as a function of left belt speed ($F(3,27)=241.50$, $p<0.01$), right belt speed ($F(3,27)=228.45$, $p<0.01$) and the interaction of both belt speeds ($F(9,81)=67.15$, $p<0.01$) were significant

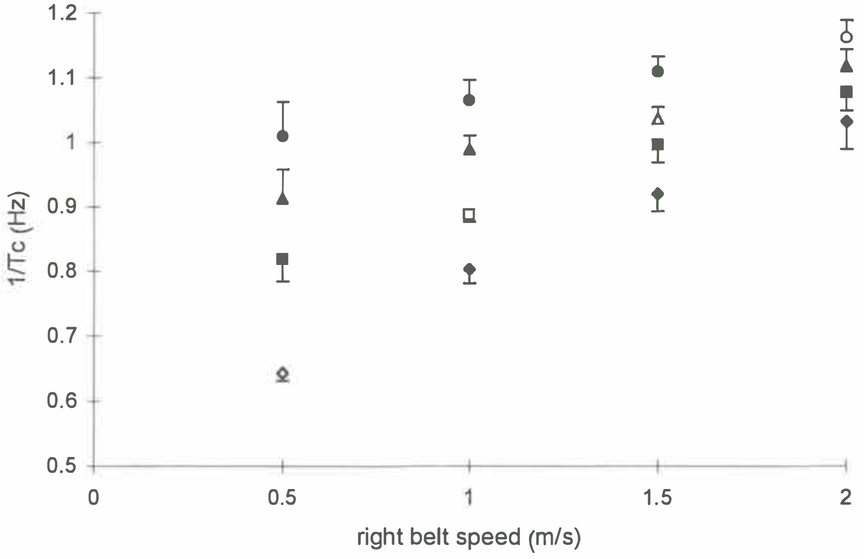

Figure 1: Stride frequency ($1/T_c$) versus right belt speed in conditions with different left belt speeds. Symbols indicate different left belt speeds; (◆) = 0.5 m/s, (■) = 1.0 m/s, (▲) = 1.5 m/s and (●) = 2.0 m/s. Normal walking conditions are indicated by open symbols. Bars indicate plus or minus one standard error.

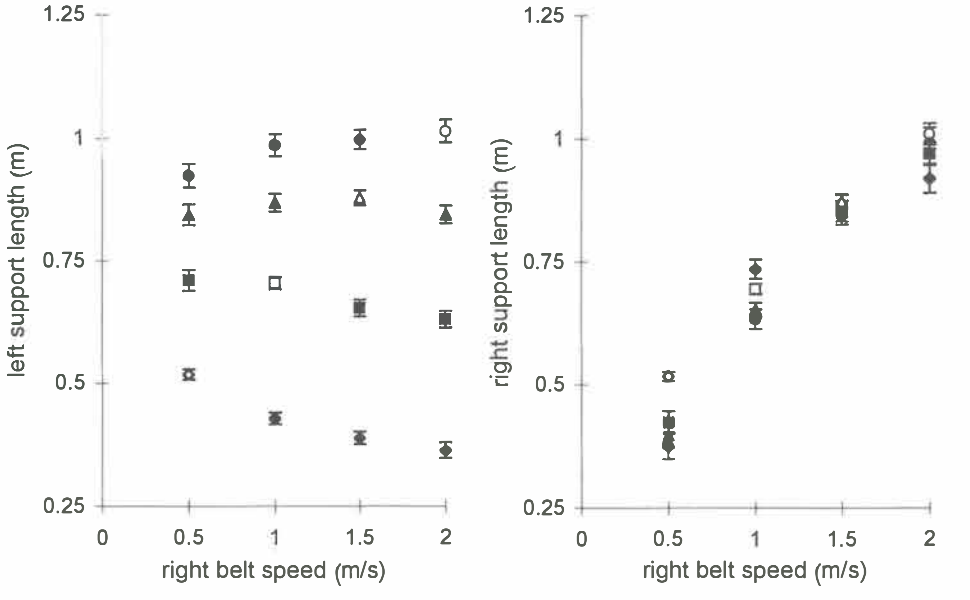

Figure 2: Left and right support lengths (support phase duration multiplied by belt speed) versus right belt speed. Symbols indicate diffent left belt speeds; (◆) = 0.5 m/s, (■) = 1.0 m/s, (▲) = 1.5 m/s and (●) = 2.0 m/s. Normal walking conditions are indicated by open symbols. Bars indicate plus or minus one standard error.

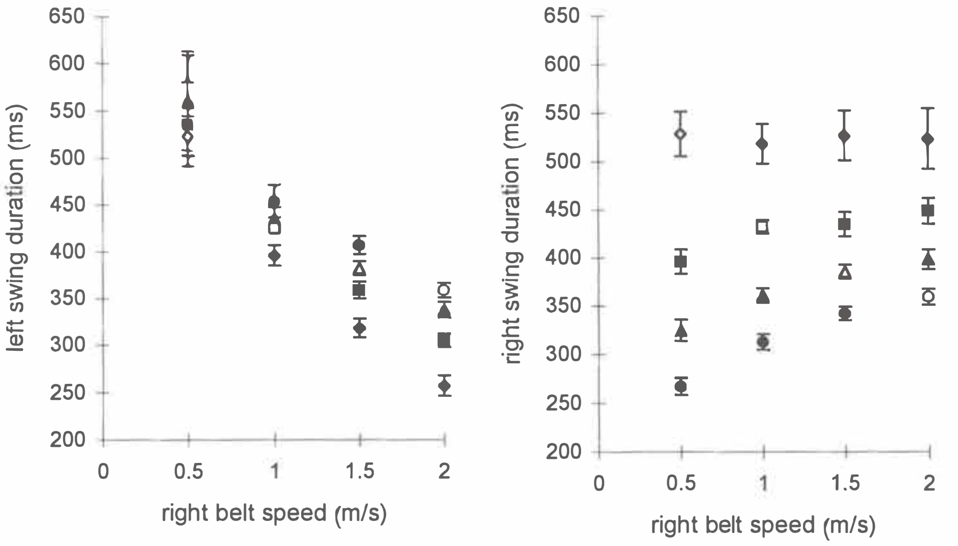

Figure 3: Swing phase durations of the left and right leg versus right belt speed. Symbols indicate diffent left belt speeds; (◆) = 0.5 m/s, (■) = 1.0 m/s, (▲) = 1.5 m/s and (●) = 2.0 m/s. Normal walking conditions are indicated by open symbols. Bars indicate plus or minus one standard error.

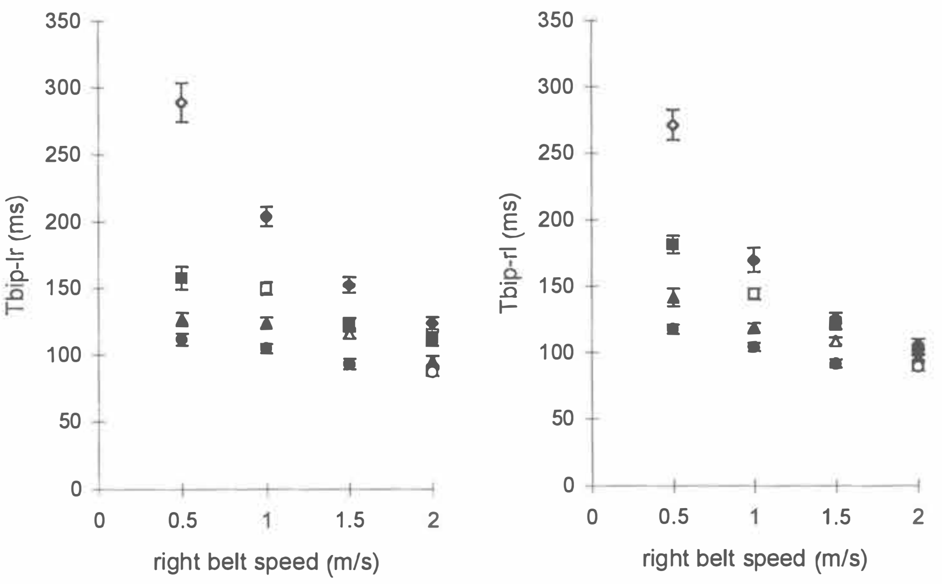

Figure 4: Durations of both double support phases versus right belt speed. Symbols indicate diffent left belt speeds; (◆) = 0.5 m/s, (■) = 1.0 m/s, (▲) = 1.5 m/s and (●) = 2.0 m/s. Normal walking conditions are indicated by open symbols. Bars indicate plus or minus one standard error.

Timing of onset / mid / end of support and swing phases versus belt speed

Phase lags were calculated between the timing of onset, mid and end of left and right support phases. In normal walking conditions, there was an alternating of approximately symmetrical left and right leg movements in such a way that any event within the movements of one limb was followed by the same event in the other limb with a phase lag of 0.5. In split-belt walking, there was no symmetry of leg movements and a phase lag of 0.5 could not be maintained for all stride cycle events. Only for the mid of both support phases was an 0.5 phase lag maintained. When phase lags were calculated between the mid of support and the mid of contra lateral swing, a coupling (the phase lag equalled 0) was observed for all conditions.

Discussion

The results of these split-belt experiments illustrate an adaptability of human locomotion which was hitherto unknown. Despite differences in left and right belt speeds up to a ratio of 1:4, all subjects maintained an alternating mode of leg movements and a more or less symmetrical duration of bipedal support phases. Thus, in all split-belt conditions and in all subjects, the movement pattern could be qualified as walking. The spatio-temporal adaptation of the stride cycle during split-belt walking in healthy adults partially agree with the results of animal experimental studies on cats walking in corresponding conditions[6]. As in animals walking with moderate left-right speed differences, our subjects adopted a stride cycle duration (i.e. a stride frequency) intermediate to the normal values for the corresponding belt speeds. The stride frequency always tended to the normal value of the fast leg. Within the stride cycle a reorganisation of support and swing durations took place which allowed for adaptations in the amplitude of leg movements for both legs. The major difference with the animal experimental results seems to be the observation that when cats were offered large speed differences the "fast" leg performed stepping movements in multiples of the "slow" leg (for example, 2:1 or 3:1)[6].

The fact that our subjects maintained an invariant phase lag of approximately 0.5 between the mid of left and right support phases (or to put it differently, the perseverance of a coupling between ipsilateral support and contralateral swing phase) can be attributed to the more or less symmetric nature of the adaptation in both double support phases. A systematic asymmetry in LR- and RL- double support phases would have resulted in a change of these phase lags and the overall disappearance of one or both double support phases would have led to a galloping[15] or running type of locomotion. That neither of the latter possibilities were observed points to a high stability in the coordinative tendency to perform alternating leg movements and to provide for recurring double support periods. This remarkable stability of the walking pattern in all split-belt conditions is also illustrated by the absence of significant increases in the calculated coefficients of variation. Although in split-belt conditions with the largest speed difference a somewhat higher variability of the stride cycle duration was present, these changes were never significant. The maintenance of an 0.5 phase lag and the symmetric adaptation of both double support phases most likely must be attributed to the demands to be fulfilled for an effective control of body equilibrium during bipedal gait. With respect to the latter, it must be kept in mind that the split-belt experiments in animals involved chronic spinalized cats of which the hindlimbs were placed on the treadmill belts whilst the fore part and the trunk of the cats were supported. Therefore the latter situation did not require postural control.

Although voluntary interactions can occur, the control of bilateral leg movements normally takes place at a subconscious level[12]. Walking in split-belt conditions did not seem to require a voluntary monitoring of the movement pattern in the adapted situation. The only aspect of movement execution which now and then needed some voluntary control was foot placement but most of our subjects managed to walk in split-belt conditions without the necessity of a visual control of foot strike. Despite large differences in left and right belt speed, our subjects automatically adapted their walking pattern and reached a rhythmic alternation of left and right steps within 10 to 20 stride cycles. The fact that some of our subjects noted that they had to readapt to the normal walking situation after walking in split-belt conditions underlines the rapid and automatic nature of the adaptation to split-belt conditions.

From our results, it can be concluded that the relative timing of support and swing phases within the stride cycle is highly modifiable. Recently[12], it was argued that the involuntary adaptation of support and swing phases within the stride cycle results from the interaction between lower level central mechanisms and the dynamics of the movement pattern of the legs during walking. Also in the present results, it appears that the automatic adaptation of leg movements seen in our subjects is a property of lower level coordinative mechanisms responding to the different conditions offered to the left and right leg. Afferent feedback obviously plays a major role in the automatic adjustments of support and swing phases of both legs. The combination of different belt speeds results in changes in both support lengths which presumably are signalled to the spinal cord by means of afferent feedback. Thus, afferent information probably contributes to adjustments in movement amplitude and timing so that stability of walking is maintained. Two kinds of proprioceptive feedback could play a role in the interlimb control during split-belt walking: 1) Information on the velocity and direction of angular displacements in the leg joints. Particularly the angular displacements in hip[17] and ankle joints seem to generate essential information about the movement pattern of the legs during walking (see also Ref. 5). 2) Information on the (un)loading of the legs[18-20]. From animal experimental work it is known that such proprioceptive feedback is able to reset the locomotor activity generated by a central pattern generator (for review see Ref. 5).

In summary: the results of these split-belt experiments again[5,9,10] confirm the existence of a high degree of similarity in basic coordinative mechanisms in animal and human locomotion. It is likely that, as in quadrupedal locomotion, separate low level coordinative mechanisms for each limb exist in human locomotion also and that feedback mechanisms play a major role in tuning these coordinative mechanisms to actual task requirements. Our results show that an automatic adaptation of leg movements takes place even in conditions as extreme as the ones offered to our subjects. The only invariance found in our data is the maintenance of an alternating pattern of leg movements. It can therefore be concluded that, contrary to the results of other studies which suggest invariant aspects in the automatic execution of patterned leg movements during locomotion[21], our and previous[12,22] results suggest a high degree of flexibility in the relative timing of leg movements during walking.

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Chapter 5: Adaptability of Leg Movements during Normal Treadmill Walking and Split-Belt Walking in Children

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Abstract

The present study addresses the question whether the coordination of leg movements during locomotion in children is similar to the coordinative patterns observed in adults. To this purpose, the spatio-temporal adaptation of leg movements to increases in walking speed and differences in left and right belt speeds was studied in 73 healthy children aged 4 to 10 years. The results of children in different age groups were compared with data of 10 healthy adults in similar conditions. A dimensional analysis of the adaptation of leg movements in children and adults suggests that physical rather than neuro-developmental maturation is the cause of age-related differences in spatio-temporal data. The main differences with adult walking patterns can be summarised as a higher intra- and inter-subject variability of spatio-temporal parameters in children. Differences in postural control may be the cause of this higher variability in children. It is concluded that despite the latter differences, the interlimb coordination and the adaptability of walking in children older than four is similar to that of adults. *Gait & Posture 1996; 4: 212-221.*

Introduction

Human walking is characterised by a high degree of flexibility. Changes in movement direction, walking speed or changes in ground surface result in an appropriate adaptation of movements within the stride cycle. Although voluntary modifications are possible[1,2], the involuntary adaptation of leg movements during changes in walking speed involves invariant changes in a number of spatio-temporal variables. The mature walking pattern is characterised by linear relationships between stride frequency, stride length and walking speed[2-6]. Increases in walking speed, and thus stride frequency and stride length, coincide with curvi-linear decreases in the duration of the stride cycle and its support and swing phases[6]. This leads to a change in the relative timing of leg movements which appears to be predominantly dependent on the amplitude of leg movements[2]. Recently, the adaptation of leg movements to walking conditions in which both legs were offered different speeds by a split-belt treadmill was described[7]. In this experiment, it was shown that despite left and right speed differences up to a ratio of 1:4, and despite the necessity of adapting all spatio-temporal parameters of the stride cycle, all subjects were able to maintain an alternating walking mode of locomotion. These results illustrate a high degree of flexibility in the relative timing of leg movements during walking and a stability in the tendency to perform an alternating type of locomotion.

In recent years, a number of studies was directed at the development of stepping patterns in infants[8-13] and in children immediately after the acquisition of independent walking[8,14-17]. These studies showed that alternating stepping patterns are already present in the youngest infants, and that the development of stepping is dependent on the parallel development of weight bearing ability, postural control, and interlimb coordination. Stepping patterns in children capable of more or less skilled independent walking deviate from adult-like walking patterns in both electro-physiological[15,18] and biomechanical[3,14,18-21] parameters of walking. Depending on the analysed parameters, the appearance of a mature walking pattern has by different authors been set at different moments on the ontological time scale (for examples, see Refs. 14, 20 and 22). This fact illustrates that it is often unclear whether the development to adult-like stepping patterns must be explained by a physical or a neuro-developmental maturation.

To investigate the effects of maturation on interlimb coordination during locomotion, the present study analyses the spatio-temporal adaptation of leg movements to increases in walking speed and differences in left and right belt speeds in children aged 4 to 10 years. The results of the children are compared with those of adults walking in similar conditions[7]. One major problem in comparing gait data of children and adults is the fact that, at least in part due to their smaller leg lengths, children walk with smaller steps and higher step frequencies than adults do[3,14,19-21]. However, assuming geometrical and dynamical similarity between children and adults, this problem can be overcome by expressing spatio-temporal data in dimensionless numbers. This method and related techniques are widely used in the technical sciences and in biology where, e.g., walking patterns of mice and dinosaurs are to be compared[23] (an extensive treatment of this subject can be found in Refs. 24, 25 and 26). Besides a dimensional analysis of spatio-temporal parameters, coefficients of variation and asymmetry will be calculated to evaluate the stability and the symmetry of the coordinative patterns in children. As studies on the walking pattern in children usually are directed at free walking, our study will also provide reference data on treadmill walking in children.

Methods

A total of 73 children (with ages ranging from 3.6 to 10.5 years, body mass ranging from 13.0 to 44.5 kg, and leg lengths ranging from 0.45 to 0.80 m) participated in the experiments. The group of children consisted of 41 boys and 32 girls. The children were divided into seven age groups. A description of the different age groups, including characteristics of a group of adult subjects used for reference data, is given in Table 1. The experimental protocol was approved by the local ethical committee, and the children's and/or parent's informed consent were obtained for the experiments.

Each child was offered all possible combinations of three belt speeds. In the four, five and six years old children these belt speeds were 0.25, 0.5 and 1 m/s. In the seven to ten- year-old children, combinations of 0.5, 1 and 1.5 m/s were offered. Three exceptions to this rule were made: two six- year-old children were offered the speed range of the older children and one seven-year-old child was offered the speed range of the younger children. The data of the children were compared with the data of 10 adult subjects walking in similar conditions, but with a wider range of speeds (0.5, 1.0, 1.5 and 2.0 m/s) (see Ref. 7). In all children, normal treadmill walking was studied first by offering the three belt speeds in an increasing order. Thereafter, the three belt speeds were offered in different combinations for the left and right leg. These split-belt conditions were offered in the same order for all children: first, all right belt speeds were higher than left belt speeds, and then the same conditions were offered vice versa. Thus, the protocol was equivalent to the one used with adults[7]. Before recording data, the children were familiarised with both treadmill walking and walking in split-belt conditions by offering two normal walking conditions and one or two split-belt conditions. During data collection, the children wore short trousers and light foot wear.

GROUP	AGE (years)	SEX (b/g)	BODY MASS (kg)	HEIGHT (m)	LEG LENGTH (m)	KFD (m)	*n*
4 YEARS	3.8 (0.25)	4b / 6g	16.7 (1.89)	1.03 (0.05)	0.50 (0.04)	0.25 (0.02)	10
5 YEARS	5.2 (0.34)	4b / 6g	19.1 (2.36)	1.09 (0.05)	0.52 (0.04)	0.26 (0.05)	10
6 YEARS	6.1 (0.29)	5b / 8g	19.9 (2.09)	1.14 (0.03)	0.55 (0.02)	0.29 (0.02)	13
7 YEARS	7.1 (0.30)	7b / 3g	24.7 (4.06)	1.24 (0.05)	0.61 (0.03)	0.32 (0.02)	10
8 YEARS	8.2 (0.28)	7b / 3g	24.5 (3.12)	1.28 (0.05)	0.65 (0.03)	0.32 (0.02)	10
9 YEARS	9.8 (0.26)	9b / 1g	29.5 (1.99)	1.35 (0.03)	0.69 (0.02)	0.36 (0.02)	10
10 YEARS	10.2 (0.27)	5b / 5g	33.7 (6.35)	1.39 (0.08)	0.73 (0.06)	0.37 (0.03)	10
ADULTS	26.7 (2.76)	all male	73.7 (10.79)	-	0.94 (0.05)	0.51 (0.03)	10

Table 1: Means and standard deviations of age, body mass, height, right leg length and knee to floor distance (KFD) for each age group. The number of subjects per age group is indicated by n. Also indicated is the distribution of boys (b) and girls (g).

Data-recording took place during 60 s of walking in a particular condition. By means of Kistler force plates which were placed under the two treadmill belts the force exerted on the treadmill could be measured for each leg (for details see Ref. 27). Force signals were amplified (floating-input amplifier; bandwidth 3-1000 Hz; sensitivity 2mV/V) and transferred on-line to a micro computer system via an A/D converter sampling at 500 Hz.

From the recorded force signals stride cycle duration (T_c) and the durations of left and right support (T_{sup}) and swing (T_{swi}) phases were determined. Stride cycle duration was defined as the time between two successive impacts of the right leg. For each leg, T_{sup} was measured as the duration of the exerted force and T_{swi} as the duration of absence of exerted force on the treadmill belt. The onset of the force signal of the right belt (i.e., the impact of the right leg) was used as a trigger for selecting stride cycles. For each child and each walking condition, at least 15 successive stride cycles were selected and used for measuring the above-mentioned temporal variables.

Further calculations and a statistical analysis (means, ranges, standard deviations, Pearsson correlation coefficients (R), (curvi)linear regression functions, *t*-tests and MANOVA) of the spatio-temporal parameters were performed by means of a statistical software package (SPSS/PC™). To correct for differences in body size, all individual measures of time (t), frequency (f) and velocity (v) were expressed in dimensionless numbers. This was done according to the following formulas:

$$\hat{t} = \frac{t}{\sqrt{\ell / g}} \qquad \hat{f} = \frac{f}{\sqrt{g / \ell}} \qquad \hat{v} = \frac{v}{\sqrt{g \ell}}$$

In these formulas, dimensionless parameters are indicated by the ^ symbol. Leg length (ℓ) (measured as the distance from the top of the right trochanter major femoris to the floor without wearing shoes) was used as an individual measure for correcting for size differences, and g is the acceleration due to gravity (9.81 m.s^{-2}). A full description of the methods for expressing the spatio-temporal data in dimensionless numbers can be found in Ref. 28. From the sum of the calculated dimensionless durations of left and right swing phases, the duration of the period of support on one leg was calculated ($T_{ssup} = T_{swi-l} + T_{swi-r}$). For all age groups, regression functions were calculated to express the relationships between different temporal parameters. It was attempted to find a best fit of the relationships between parameters by using different regression functions. A coefficient of variation ($VC\text{-}T_c = 100 * $ (standard deviation of T_c / T_c)) was calculated to express the individual variability in stride cycle duration. Furthermore, an asymmetry coefficient was calculated to quantify differences in the durations of left and right support phases ($Ac_{sup} = 100 * (T_{sup} / T_{sup})$). All calculations were made for each child and each condition.

Results

Although all children were able to perform unsupported stepping movements during all treadmill walking conditions, some children were uncertain and needed some support during the start of some walking conditions. This occurred particularly in the youngest children and usually in split-belt conditions. In these children, support was provided by lightly touching the rail around the treadmill. Care was taken that children did not support too long and too heavily. The selection of stride cycles for an analysis of the walking pattern of children was always restricted to those parts of the recording period where the children did not use additional support. In the following sections, the results of normal treadmill walking will be presented first. Subsequently, the adaptation of leg movements to increases in belt speed will be analysed by means of comparing the trends in calculated dimensionless variables of all age groups. Finally, an analysis of the adaptation of the temporal parameters during split-belt locomotion will be made.

Normal treadmill walking

In Table 2, the group mean values of the measured variables are listed for all age groups. The temporal variables (T_c, T_{sup} and T_{swi}) decreased with speed in all age groups. Although the differences decreased with increasing age, the adults always had a longer duration of T_c, T_{sup} and T_{swi} than children walking at the same speed. Despite the longer durations of these temporal variables the standard deviations (indicating inter subject differences within groups) most often were lower in the adults.

Apart from these differences in absolute values and standard deviations, a clear distinction between adults and all children groups can be made when observing the variability of the stride cycle duration ($VC\text{-}T_c$) and the asymmetry in left and right support durations (AC_{sup}). In all conditions, the values of the adults indicated a lower variability and a higher symmetry within the stride cycle. The variability of the stride cycle decreased with increasing belt speed in all age groups. The symmetry of leg movements increased with speed in all groups but it increased substantially more in the youngest children. These effects of belt speed and age group were evaluated by means of a MANOVA two-factor design. This revealed significant main effects for both factors, but no significant interaction effect. Univariate F-tests showed significant effects for speed on $VC\text{-}T_c$ ($F(2,193)=29.29$, $p<0.001$) and AC_{sup} ($F(2,193)=5.20$, $p<0.01$), and for age group on $VC\text{-}T_c$ ($F(7,193)=11.02$, $p<0.001$) and AC_{sup} ($F(7,193)=2.68$, $p<0.05$). The effects of belt speed and age group remained when the adult group was excluded from the analysis.

GROUP	SPEED (m/s)	TC (ms)	VC-TC (%)	Tsup (ms)	ACsup (%)	Tswi (ms)	*n*
4 YEARS	0.25	1232.2 (263.8)	7.0 (1.89)	790.0 (192.1)	92.8 (5.64)	452.7 (95.4)	10
	0.50	1034.5 (137.6)	6.7 (2.33)	624.9 (75.1)	96.5 (3.27)	409.9 (73.3)	10
	1.00	760.3 (47.6)	5.2 (2.26)	442.0 (26.7)	96.9 (1.88)	318.5 (25.4)	10
5 YEARS	0.25	1240.8 (190.4)	7.3 (2.20)	766.5 (103.0)	91.5 (4.19)	477.8 (102.2)	10
	0.50	1023.4 (77.3)	6.0 (1.29)	618.4 (39.3)	96.4 (2.56)	404.1 (49.6)	10
	1.00	811.2 (62.0)	3.8 (1.15)	473.5 (33.8)	97.2 (1.72)	338.0 (36.1)	10
6 YEARS	0.25	1254.8 (130.6)	6.7 (2.29)	779.9 (88.2)	94.7 (4.75)	473.3 (58.9)	11
	0.50	1035.9 (103.2)	5.6 (1.36)	626.6 (58.2)	96.6 (2.57)	407.9 (49.9)	13
	1.00	803.6 (55.1)	4.6 (1.69)	469.1 (37.8)	96.8 (2.00)	335.5 (25.9)	13
7 YEARS	0.50	1120.3 (120.8)	5.2 (1.33)	674.7 (74.5)	94.4 (3.48)	444.4 (58.9)	10
	1.00	870.8 (53.0)	4.0 (1.24)	504.3 (31.3)	97.7 (1.65)	367.1 (31.1)	10
	1.50	742.3 (35.1)	2.9 (0.76)	416.9 (20.8)	98.2 (1.03)	326.3 (18.1)	9
8 YEARS	0.50	1159.5 (165.6)	5.4 (0.87)	687.5 (91.2)	96.7 (1.78)	473.3 (80.0)	10
	1.00	898.0 (58.5)	3.7 (1.27)	516.0 (33.6)	97.8 (1.19)	381.9 (28.0)	10
	1.50	755.0 (38.3)	3.5 (1.22)	426.0 (24.3)	98.2 (1.18)	329.0 (15.0)	10
9 YEARS	0.50	1217.3 (175.4)	5.1 (1.13)	733.7 (104.3)	96.0 (1.95)	485.2 (79.0)	10
	1.00	939.5 (86.1)	3.0 (0.70)	550.9 (42.1)	97.4 (1.81)	389.8 (51.4)	10
	1.50	785.9 (45.1)	2.8 (0.89)	452.3 (30.6)	97.7 (1.25)	333.4 (21.3)	10
10 YEARS	0.50	1147.1 (134.3)	4.3 (0.93)	685.5 (77.7)	97.5 (1.58)	460.9 (62.8)	10
	1.00	925.2 (123.1)	3.4 (1.38)	542.8 (72.2)	97.6 (1.83)	381.8 (53.4)	10
	1.50	765.8 (81.6)	2.4 (0.77)	436.9 (49.7)	98.1 (1.17)	329.4 (34.3)	10
ADULTS	0.50	1559.8 (89.2)	2.4 (0.55)	1032.9 (52.8)	98.3 (1.42)	528.2 (67.9)	9
	1.00	1127.4 (47.7)	1.7 (0.33)	694.8 (33.6)	98.4 (1.11)	432.3 (21.7)	10
	1.50	967.5 (51.4)	1.3 (0.35)	582.4 (31.2)	98.8 (0.77)	385.5 (24.0)	10

Table 2: Means and standard deviations for the measured temporal variables (T_c, T_{sup} and T_{swi}) and calculated coefficients of variation (VC-T_c) and asymmetry (AC_{sup}). The number of subjects per age group and speed condition is indicated by *n*.

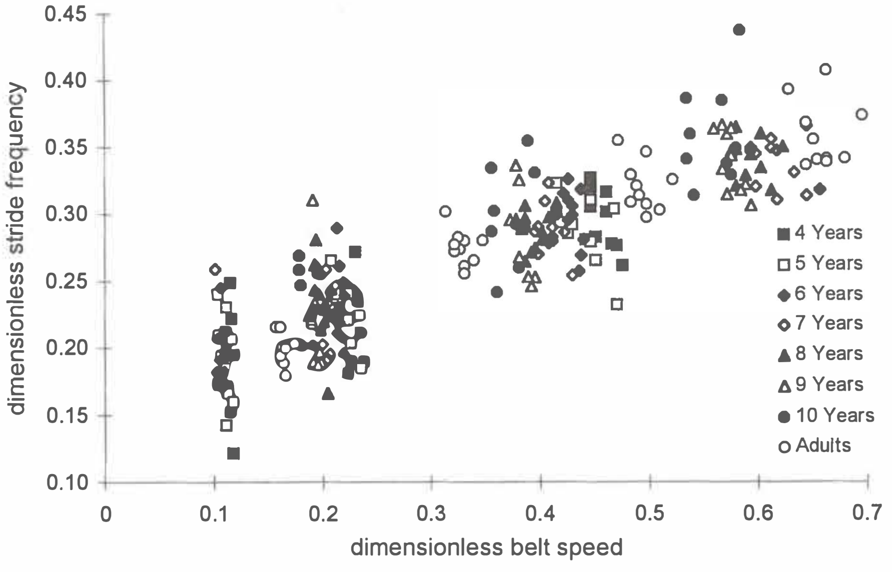

Figure 1: Dimensionless stride frequency versus dimensionless belt speed during normal treadmill walking. The values in the figure represent the individual data of all subjects. Legends in the figure indicate the symbols used for different age groups.

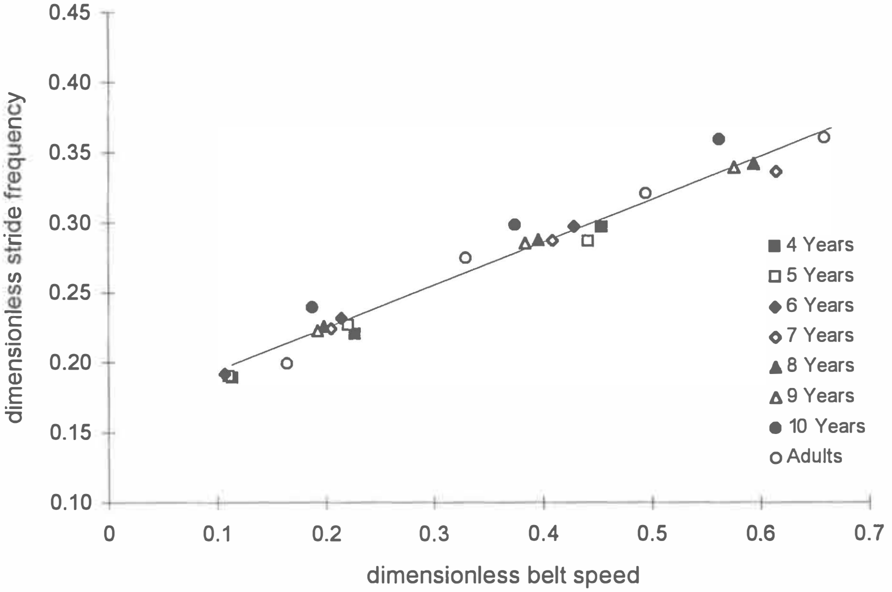

Figure 2: Dimensionless stride frequency versus dimensionless belt speed during normal treadmill walking. The values in the figure represent the means per age group and speed condition. Legends in the figure indicate the symbols used for different age groups. The solid line represents the linear regression function for all subjects (see text).

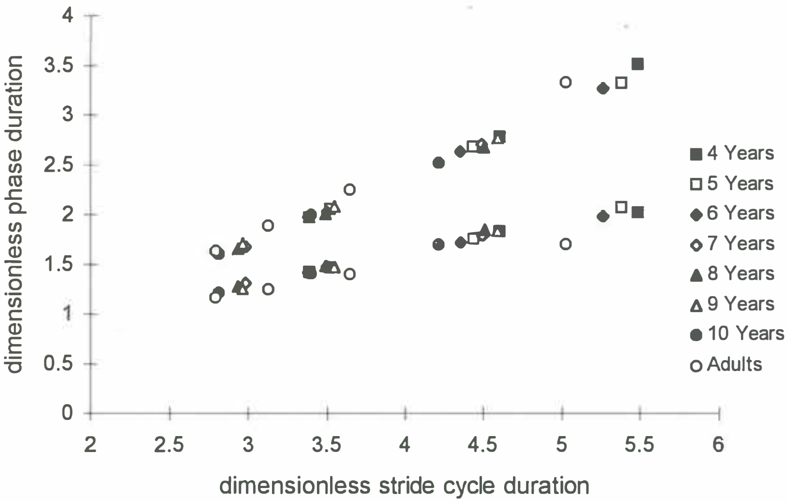

Figure 3: Dimensionless durations of the support and swing phases versus dimensionless cycle duration during normal treadmill walking. Values in the figure represent means per age group and speed condition. Lower values indicate swing durations, and the upper values indicate support durations. Legends in the figure indicate the symbols used for different age groups.

A dimensional analysis of the adaptation of leg movements to increases in speed

After correcting the data for differences in body size, subjects in all age groups showed a similar increase in frequency of leg movements with belt speed. Despite the already noted higher inter-subject variability in children (see Table 2), this can be seen in the individual data points of all subjects (figure 1), but it becomes even more clear when the group means for dimensionless stride frequency are plotted against dimensionless walking speed (figure 2).

For the whole group of subjects (children and adults), the change in stride frequency with increasing belt speed (seen in figures 1 and 2) could be described by a linear function:

$$1/T_c = 0.304 * V + 0.164 \; (R^2 = 0.79, p < 0.0001).$$

When other regression models were tried to obtain a better (curvi-linear) fit, the quality of fit (as indicated by R^2) did not improve. The calculated functions per age group showed that the best linear fit was obtained in the adults (R^2 values were 0.70 (4 years), 0.67 (5 years), 0.81 (6 years), 0.79 (7 years), 0.83 (8 years), 0.69 (9 years), 0.73 (10 years) and 0.88 (adults)). This difference between adults and children was also reflected in the individual data of children; contrary to the adults, not all children increased their speed by a linear increase in stride frequency. Particularly in the 4 and 5 years old, some children had a curvi-linear increase of stride frequency with increasing belt speed.

When the relationship between dimensionless stride cycle duration and dimensionless support and swing durations was studied, it could be seen (figure 3) that the group mean values in these parameters did not differ between the groups of children and that only a small difference existed between children and adults. In all age groups, a good fit was obtained when linear functions were calculated to express the relationship between support and swing duration with stride cycle duration. For the relationship between support duration and cycle duration, the calculated R^2 values ranged from 0.96 (in the 5-year-olds) to 0.99 (in the 8-year-olds). The adults had an intermediate value ($R^2 = 0.98$). Similar coefficients for the linear functions were obtained in all groups of children. In the adults, the increase in support duration was somewhat steeper than in children. The regression functions were:

$T_{sup} = 0.648 * T_c - 0.217$ ($R^2 = 0.97$, $p < 0.0001$) (for all children) and

$T_{sup} = 0.750 * T_c - 0.458$ ($R^2 = 0.98$, $p < 0.0001$) (for the adults).

By means of linear regression functions, the duration of single support (T_{ssup}) was predicted for three stride cycle durations (4.5, 4, 3.5) for each age group separately. These dimensionless stride cycle durations corresponded to dimensionless speeds of approximately 0.20, 0.30, 0.40. Subsequently, the predicted T_{ssup} was expressed as a percentage of T_c (see Table 3). The predictions show that, at the same speed, adults always had a lower single support percentage.

GROUP	Tc = 4.5	Tc = 4.0	Tc = 3.5
4 YEARS	78.5	79.8	81.5
5 YEARS	79.9	80.0	80.1
6 YEARS	78.5	79.9	81.6
7 YEARS	80.8	82.1	83.8
8 YEARS	83.2	84.0	85.0
9 YEARS	82.5	82.9	83.4
10 YEARS	81.5	82.2	83.1
ADULTS	70.4	73.0	76.2

Table 3: Predicted values for the relative duration of support on one leg (T_{ssup} expressed as a percentage of T_c). Per age group, predictions were made for three dimensionless cycle durations (4.5 / 4.0 / 3.5).

Because of the larger variability in spatio-temporal parameters between children (see figure 1, and see the standard deviations of T_c, T_{sup} and T_{swi} in Table 2), our data were studied further to indicate factors which contribute to inter-subject differences in dimensionless stride frequency and dimensionless support and swing duration. This was done by means of an analysis of the correlation coefficients between different relevant parameters, and by a comparison of group means (*t*-tests). Correlation coefficients were calculated within groups and over the whole group of subjects. No significant effects could be shown for age, gender, body mass, or body mass normalised to group mean body mass.

The adaptation of the stride cycle during split-belt locomotion

Despite the fact that some of the younger children often needed support during the start of a split-belt condition, all children were able to adapt their leg movements to the offered speed differences. As the adults, the children automatically reached a rhythmic alternation of left and right steps. Whereas all adult subjects performed regular stepping movements in the split-belt conditions, the children showed a higher degree of variability in stepping movements. This could be observed during the measurements and this was also clearly present in the calculated coefficients of variation. Despite their higher variability, the children also adapted to all conditions offered. This adaptation involved changes in all analysed stride cycle parameters.

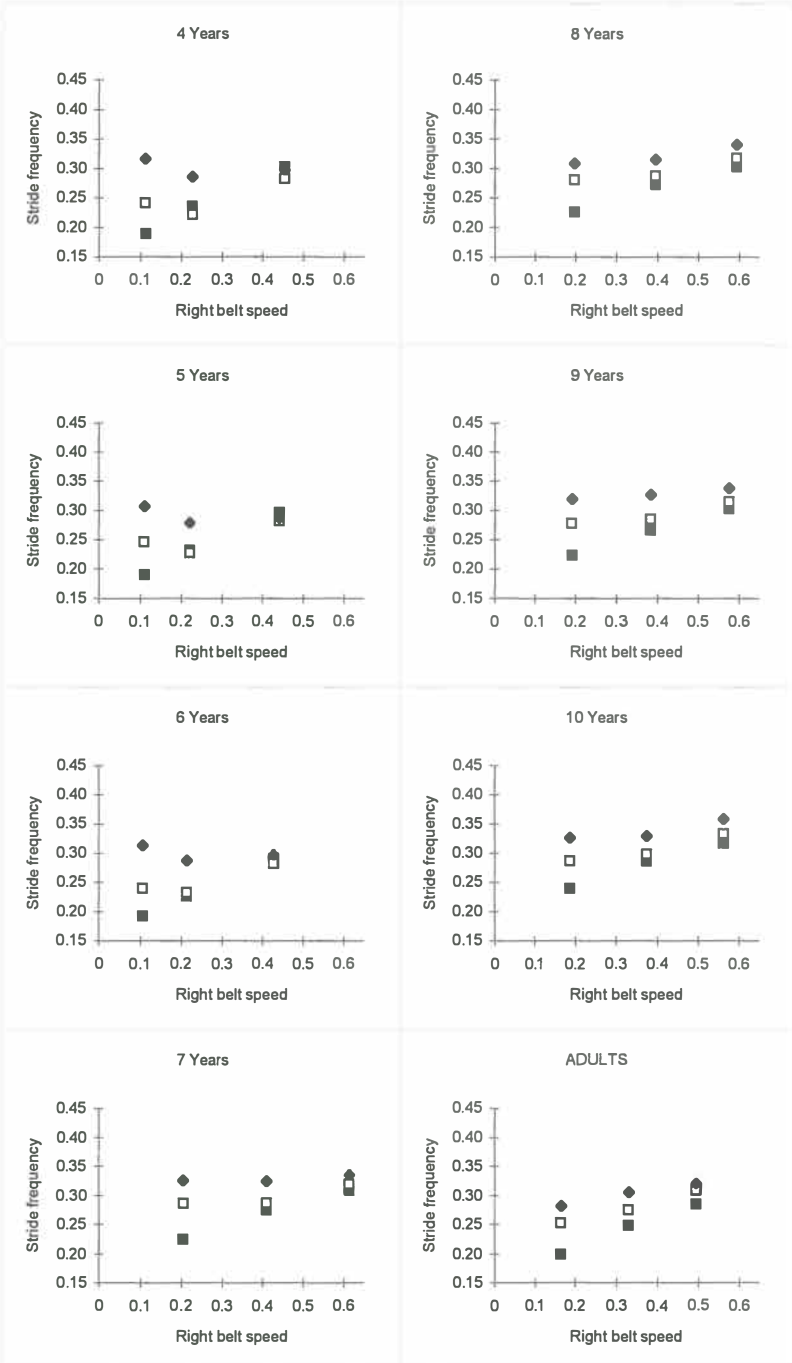

Figure 4: Dimensionless stride frequency versus dimensionless right belt speed during all split-belt conditions. Each figure represents the changes in stride frequency for the particular age group indicated in the text above the figure. The symbols indicate different left belt speeds: (■) = lowest left belt speed, (□) = middle left belt speed and (◆) = highest left belt speed.

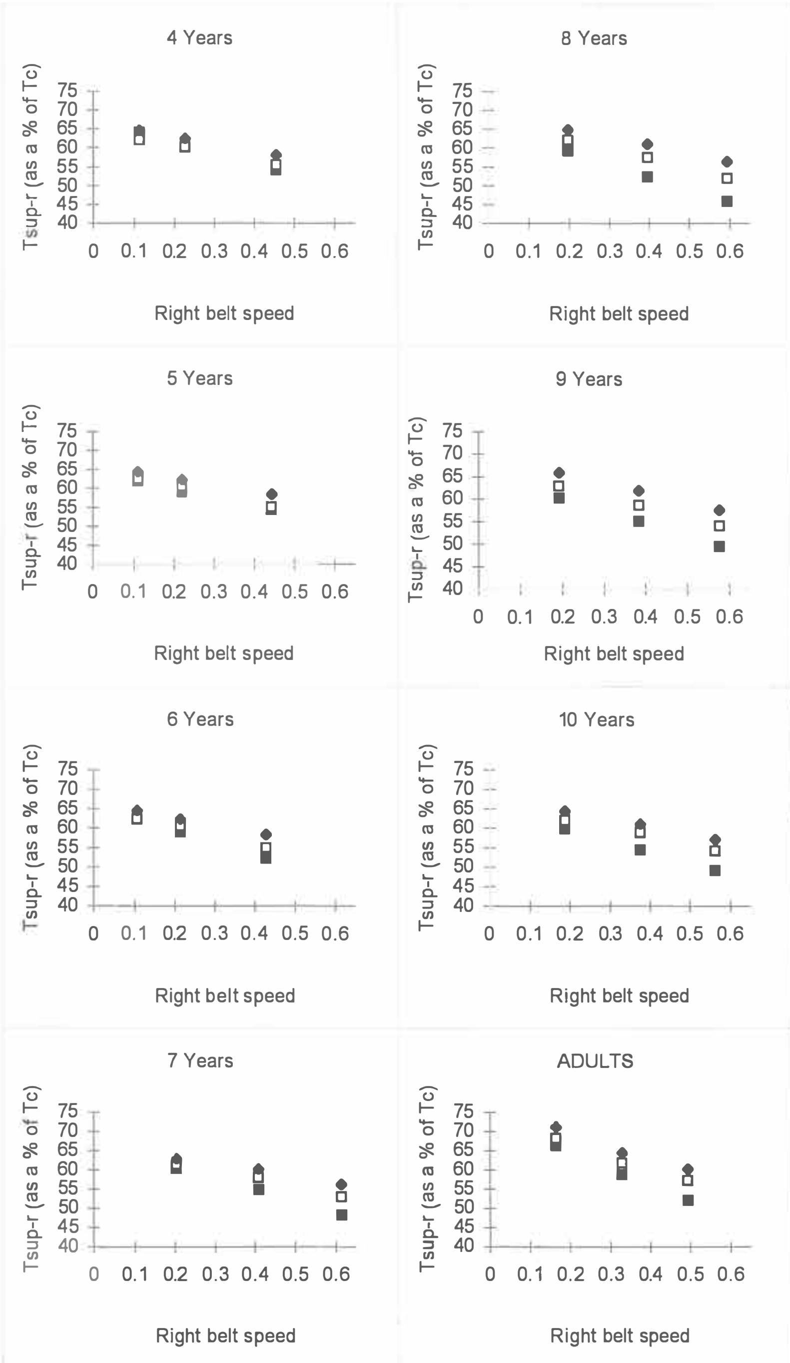

Figure 5: The relative duration of the right support phase (T_{sup} as a percentage of T_c) versus dimensionless right belt speed during all split-belt conditions. Each figure represents the changes for the particular age group indicated in the text above the figure. The symbols used for indicating different left belt speeds are as defined in figure 4.

Generally, the adaptation to split-belt conditions can be described as follows: in both legs, a stride frequency was adopted which tended to the frequency during normal walking at a speed corresponding to the faster belt speed (figure 4). An asymmetric adaptation in the durations of support and swing phases within the stride cycle coincided with different amplitudes of leg movements in the left and right leg. The always larger amplitude of the leg on the faster belt was allowed for by an elongation of the swing phase of this leg (and correspondingly the support duration of the contralateral "slow" leg). Vice versa, the support duration in the "fast" leg and the swing duration in the "slow" leg were decreased. This asymmetric adaptation in the durations of support and swing phases within the stride cycle resulted in changes in the relative timing of support (figure 5) and swing phases which depended on both left and right belt speed.

Although the adaptation of leg movements corresponded to the above-mentioned strategy in all children, differences were observed between the younger age groups (4, 5 and 6 years old) and the older age groups. In figure 4, it can be seen that, whereas the older children and the adults modified their stride frequency to a value intermediate to their frequency at corresponding normal speeds (with a tendency to the normal value for the fast speed, however), the younger children showed a less systematic adaptation to the split-belt conditions. It was observed that in the younger children the stride frequency in a particular split-belt condition sometimes was higher than the stride frequency measured during normal walking at a speed corresponding to the highest of both belt speeds. This can be seen in the four, five and six years old children, in the conditions where the lowest right belt speed was combined with other left belt speeds (figure 4). However, in this case, one has to bear in mind that despite the fact that the variables are expressed in dimensionless numbers there still exists a small difference between the speed range offered to the 4, 5 and 6-year-olds and the one offered to the 7, 8, 9 and 10-year-olds. The lowest belt speed for the former groups was clearly lower than that of the latter groups and even somewhat lower than the lowest dimensionless belt speed for the adults.

The reorganisation of the durations of the support and swing phases also differed slightly between age groups. This reorganisation of phase durations was always directed at enabling the leg on the faster moving treadmill belt to make a larger amplitude but, although this strategy was maintained in all age groups, the range in which support (and swing) durations were adapted was larger in the older children and the adults (figure 5).

Discussion

The aim of this study was an evaluation of interlimb coordination during locomotion in children by means of an analysis of the adaptability of leg movements during treadmill walking at different speeds and during walking in split-belt conditions. To this purpose, temporal parameters of walking in different age groups were analysed and compared to the results of adults walking in similar conditions. The following discussion of results will concentrate on the comparison with adult walking patterns. When differences are observed, possible factors contributing to these differences will be discussed.

The absolute values for the analysed parameters of normal treadmill walking confirm the known differences between adults and children. At the same walking speed, children walk with higher stride frequencies than the adults do (see Table 2 and see for example Ref. 3, 14, 18-21). Despite differences in absolute values, the durations of the stride cycle and the swing and support phases changed with speed in a similar manner as in adults. As could be expected, the differences with adult values become less with increasing age. From the calculated dimensionless data, it becomes clear that existing differences in temporal parameters between adults and children as listed in Table 2 do not remain when the data of children and adults are corrected for differences in body size. After such a normalisation, the adaptation of stride frequency, support and swing phases to increases in speed show similar linear trends in adults and children. However, a somewhat steeper increase in the duration of the support phase with increasing cycle duration could be noted in the adult group. This, and the calculated estimations of the duration of the period of support on one leg, implies that the adults have longer periods of double support than children. In earlier gait studies, the relative duration of the time in which body weight is supported by only one leg has been mentioned as one of the determinants of mature gait[14,18]. It was hypothesised that a longer duration of support on one leg reflects the ability of a subject to maintain balance during walking. Our data show that for the same dimensionless duration of the stride cycle, and thus for a similar speed, the adults actually have a lower single support percentage than the children. Thus, it can be concluded that the lower single support percentage for children found in earlier studies[14,18], and also the differences in absolute values of spatio-temporal parameters which are normally found in studies of walking children, reflect physical maturation and not the maturation of neural coordinative mechanisms.

When individual data were considered, it was observed that especially in the younger children the relationship between walking speed and stride frequency differed from that in adults. Whereas all adult subjects adapted to increases in treadmill speed by a linear increase in stride frequency, these children often showed a less systematic increase. This contributed to an overall higher inter-subject variability of temporal parameters in all groups of children. A second observation in the individual data of children and adults was a higher intra-individual variability in cycle duration, and a higher asymmetry within the stride cycle in children. In fact, these two parameters indicate the most significant differences with adult values which were found in this study. In all conditions the values of the adults indicate a lower variability and a higher symmetry within the stride cycle.

Although all subjects maintained a walking mode of locomotion in split-belt conditions, the adaptation of temporal parameters of the stride cycle was slightly different in the younger age groups. Whereas adults always modified their stride frequency to a value intermediate to the stride frequencies in corresponding normal walking conditions, some of the children showed a less systematic adaptation. In all age groups, the support and swing durations were adapted asymmetrically to allow for a difference in the amplitude of left and right leg movements. The support duration of the leg on the slower moving treadmill belt was increased together with the duration of the swing phase of the leg on the faster moving treadmill belt (vice versa, a decrease in swing duration of the slow leg and support duration of the fast leg was seen). When comparing the adaptation in children with the results in adults, it can be concluded that also these adaptations were done in a similar manner. However, the adaptation to split-belt conditions again appears to be less systematic in the youngest children and the range in which support (and swing) durations were adapted was largest in the older children and the adults.

From this discussion of our data, it can be concluded that the present study shows that interlimb coordination of leg movements within the stride cycle is similar in children and adults and that, contrary to earlier suggestions[14,20], differences in body size rather than neuro-developmental maturation are the cause of differences in absolute values of spatio-temporal data. The main differences observed in this study can be summarised as a higher asymmetry, and a higher intra- and inter-subject variability in children. The latter is obviously caused by the less systematic adaptation of temporal parameters in children. Whether these differences are due to the children being at unease during treadmill walking is unclear. To our knowledge, studies on the walking pattern of children during free walking do not report measures on the individual variability in movement patterns. Therefore, a direct comparison with free walking cannot be made. But, despite this possible cause of differences, we think that other factors contribute to the higher variability of spatio-temporal parameters. Studies of the early development of stepping patterns show that for stepping to develop, the postural task must be mastered first[8,9,13]. A flexibility of locomotion, as required during the adaptation of leg movements to increases in speed and to split-belt conditions, obviously demands an optimal control of dynamic equilibrium. It can be questioned whether the children in our study already are able to do so. Depending on the postural task and the analysed parameters several age-dependent deviations from mature motor behaviour have been found in children who already are capable of independent walking[29-31]. Other studies indicate that postural control during locomotion in 6 years old and 8 years old children show differences with adult strategies[32-34]. Deviations in postural control like those in the aforementioned studies can, on the one hand, be attributed to the fact that during postural control children are at disadvantage because of their smaller lengths[30]. On the other hand, children may not have developed the postural mechanisms which are necessary for skilled postural performance. Regardless which of these two factors is decisive, we suggest that differences in postural control play a role in the observed differences in adaptation between adults and in children in our data.

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Chapter 6: Displacement of the pelvis during human walking: experimental data and model predictions

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Abstract

Displacements of the pelvis during treadmill walking were studied in dependence of walking speed, stride frequency, and stride length. Displacement curves per stride cycle were described by means of harmonic analysis. Simple mechanical, or geometrical, models of the body's centre of mass trajectory during walking were used to predict amplitude and timing of pelvic displacements. As predicted by inverted pendulum models, the amplitude of pelvic displacement in vertical direction depended on stride length. Anterior-posterior displacements were predicted by assuming equal maxima of potential and forward kinetic energy of the body. Predictions for left-right displacements were based upon a model that assumed a constant stride width and a sinusoidal movement pattern. In agreement with this model, the amplitude of pelvic displacement in left-right direction depended on stride frequency. The presented models give insight into mechanical mechanisms which determine the pelvic trajectory during walking. The presented data may be useful in the clinical evaluation of gait disturbances. *Gait & Posture (in Press).*

Introduction

The general goal of any form of animal locomotion can be defined as the displacement of the body's centre of mass (COM). In terrestrial locomotion, this is achieved by limb movements generating propulsive forces which result in a displacement of the trunk[1]. In human bipedal locomotion, any trunk displacement is achieved by means of lower limb movements which primarily result in a forward displacement of the pelvis which is the supportive structure for the head, arm, and trunk (HAT) segments. During walking, the pelvis moves forward and downward with each step, and it subsequently moves upward when body weight is transferred from one leg to the other. Thus, with each step, the pelvis moves up and down and, relative to the mean forward velocity, it displays speed fluctuations in the direction of progression[2]. The weight transfer from one supporting leg to the other also coincides with cyclic left-right movements of the pelvis. Due to an active stabilisation of the HAT[3,4], the movement pattern of the HAT largely follows that of the pelvis[5,6]. Thus, the basic pattern of displacements of both pelvis and HAT is a movement pattern which, during one stride, displays two oscillations along the vertical axis, two oscillations in anterior-posterior direction, and one oscillation in left-right direction.

It is well known that shape and amplitude of the pelvic and trunk displacements vary with speed of locomotion[5-7], but as the above discussion already indicates, these variations may be ultimately determined by step length and stride frequency. When the walking pattern of an individual is studied, spontaneously chosen combinations of stride length and frequency are observed which usually have a high degree of repeatability[2,8]. However, humans are capable of voluntary adaptation of stepping movements (e.g. Ref. 8, 9), and in case of pathology, deviations from the normally observed combinations of stride length and frequency in healthy subjects may be seen as a result of the specific pathologic condition. It is very likely that such deviations will result in changes of the normal displacement pattern of the pelvis and HAT. Thus, although displacements of the pelvis, trunk and head have been studied with varying speeds of locomotion, it may be more appropriate to study them in their dependency of both stride length and stride frequency.

In order to get more insight into the mechanisms which determine the trajectory of pelvis and HAT during walking, the present study was directed at an analysis of pelvic displacement during walking. As the COM lies within the pelvis (in anatomical position the COM is close to a position in front of the second sacral vertebra), models of the COM trajectory during walking can possibly be used to predict patterns of pelvic displacement. It will therefore be attempted to make predictions of the amplitudes and timing of the oscillations of the pelvis by means of simple models of the COM trajectory. To this purpose, it will be assumed that the displacement patterns of COM and pelvis can be approximated by sinusoidal functions. This has as an advantage that their trajectories can be modelled more easily. A harmonic analysis, as already applied by Capozzo[5,10], will be used for describing patterns of pelvic displacement, and for determining the validity of the assumption that patterns of pelvic displacement to a large part correspond to sinusoidal displacement patterns. This method uses the stereotypical movement pattern of the pelvis and trunk to define an intrinsic and extrinsic pattern of upper body movement. Ideally, the pelvis and HAT movements should be characterised by a perfect symmetry of displacement with respect to the anatomical planes, and be cyclo-stationary with a periodicity of one stride in left-right direction and half a stride (one step) in vertical and anterior-posterior directions. A frequency decomposition of the displacement functions of pelvis and HAT in vertical and anterior-posterior directions should in this ideal case consist of even harmonics only, and similarly, the movements in left-right direction should consist of odd harmonics only. These are the intrinsic harmonics of locomotion. The first intrinsic harmonic of the displacement function for a particular movement direction indicates the basic pattern of displacement during walking, it is therefore called the fundamental intrinsic harmonic. All harmonics other than intrinsic harmonics are called extrinsic harmonics, they may be regarded as indicative for an asymmetry in the act of locomotion.

The objective of the present study is to evaluate the extent to which patterns of pelvic displacement of healthy subjects can be predicted by simple models of the COM trajectory. To this purpose we present a new model for left-right movements of the COM during walking, and we use variants of existing models that have not yet been systematically applied for the prediction of vertical and anterior-posterior displacement patterns of the pelvis during walking. A second purpose of our study is to present normative data for the amplitudes and timing of displacements of the pelvis at a broad range of speeds, stride lengths, and stride frequencies.

Methods

General procedures and experimental conditions

Ten healthy male subjects volunteered to participate in the experiments. Their ages ranged from 21 to 33 years (mean age 25.4 years). Leg length was measured as the distance from the top of trochanter major femoris to the floor, it ranged from 0.92 to 1.02 m (mean 0.98 m). The subjects' heights ranged from 1.72 to 1.89 m, mean height was 1.82 m. Body mass ranged from 57 to 92.5 kg (mean body mass 72.7 kg).

Subjects were asked to walk on a treadmill at 6 different speeds in an increasing order (0.5, 0.75, 1.0, 1.25, 1.5 and 1.75 m/s). At each treadmill belt speed, data were collected in three conditions: first, during normal walking at the particular speed; then with the subject making larger steps than his usual step length; and lastly, with the subject making smaller steps than his usual step length. Subsequently, the next speed condition was offered. All subjects were given the opportunity to practice treadmill walking (and modulating their step length) before the experiments started. Subjects were instructed to look straight ahead and to avoid changing position in left-right or anterior-posterior direction on the treadmill during data recording.

Data recording

In each condition, data were collected during 25 s of walking. Data acquisition involved measuring the signals of eight force transducers placed under the surface of the treadmill and a kinematic acquisition, both by means of an Elite Movement Analysis System sampling at 50 Hz. The force transducers were placed under a left and a right supporting surface which were physically separated by a distance of 0.01 m. The treadmill belt moved over these supporting surfaces. Kinematic data were obtained by attaching small light reflecting markers on different anatomical locations and recording the marker positions by means of two Elite cameras. For the present study, only use was made of two markers placed on the left and right Spina Iliaca Posterior Superior (SIPS). The two cameras were installed behind the walking subjects. The orientation of the reference frame was conform the ISB recommendations for reporting of kinematic data[11]. Consequently, increasing x values correspond to a forward movement, increasing y values to an upward movement, and an increase in z values corresponds to a displacement to the right, relative to the subjects' line of progression.

Data analysis

The force signals obtained from the transducers under each walking surface were summed to obtain an approximation of the vertical components of the left and right ground reaction forces. These force traces were used to analyse the temporal characteristics of the stride cycle. The onset of an increase in the force signal of the right leg was used as a trigger for determining the beginning and end of a stride cycle. Thus, it was possible to determine the duration (T_c) of each recorded stride cycle. Durations of support phases were measured as the period during which a leg exerted force on the treadmill. Swing duration was measured as the period in which no force was exerted. Durations of both double support phases were measured as the durations of simultaneous support of the left and right leg. Stride length (S_c), and the displacement during single (S_{sup}) and double (S_{bip}) support, were estimated by multiplying the duration of the stride cycle (T_c), single support, or double support, by treadmill belt speed, respectively.

From the recorded kinematic signals, the 3D co-ordinates of the left and right SIPS were used to calculate the 3D trajectory of a point on the pelvis located halfway the line which connects the left and right SIPS. For the purposes of the present study, these kinematic data were left unfiltered before further processing. The recorded force signals were used for defining the beginning and end of the kinematic signal within a stride cycle. The next step in analysing our data was a harmonic analysis of the kinematic signals within each recorded stride cycle. After detrending the kinematic signal over a stride cycle, the parameters of the first ten harmonics were estimated as a Fourier-series. For displacements in anterior-posterior direction the following Fourier-series was calculated:

$$x(t) = a_{x0} + \sum_{k=1}^{10} a_{xk} \cos\left(2\pi\, k f_c t + \varphi_{xk}\right) \qquad (1)$$

a_{x0} represents the mean position over the stride cycle; a_{xk} and φ_{xk} are the amplitude and phase of the k-th harmonic, respectively; and f_c is stride frequency. Stride frequency was determined as the inverse of the duration of the particular stride cycle under consideration. Thus, for each analysed stride cycle its stride duration was taken as the fundamental period of the harmonic analysis. Fluctuations in stride frequency during the recording period could, therefore, not influence the harmonic spectrum. The subscript x in function (1) can be replaced by y or z to obtain the functions for movements in vertical or left-right direction. To conform to a usual mode of presentation in studies of locomotion, the values for phase will be presented as percentages of the stride cycle instead of degrees or radians.

Apart from the harmonic analysis of the displacement curves per stride cycle, amplitudes of displacement were also determined by calculating the difference between minimum and maximum displacement for each stride cycle. In all conditions, at least ten stride cycles were analysed and used for the calculation of individual mean values and standard deviations of spatio-temporal parameters, and the parameters of the first ten harmonics of each displacement function. To obtain measures of the intra-individual variability of the walking pattern in a particular condition, the standard deviations of spatio-temporal parameters and harmonic amplitudes were expressed as a percentage of their mean values in that particular condition. These coefficients of variation, the individual mean spatio-temporal parameters, and the individual means of the parameters of each displacement function were all used for calculating group mean values. The subsequent data analysis was directed at establishing the degree of correspondence of individual and group mean data to the models which will be described in the next sections. Linear regression functions, between the amplitudes of displacement as predicted by a model, and as described by the fundamental intrinsic harmonic of the displacement function, were calculated to obtain an indication of the validity of the models.

Predictions of the displacements of the pelvis

Displacement in vertical direction

The vertical movement of the COM during walking can be modelled as resulting from a compass gait type during single support and an approximately horizontal trajectory during the periods of double support[12]. When this inverted pendulum model is taken as a starting point for estimating the vertical movement of the pelvis ($\Delta\, y$), a prediction can be made based upon the geometric relations shown in figure 1 (upper part):

$$\Delta\, y_1 = \ell\left(1 - \cos\theta\right) \tag{2}$$

In equation (2) and figure 1, $\Delta\, y_1$ represents the vertical displacement of the pelvis during single support, ℓ is leg length, and θ is the angle between the vertical axis and the leg.

As the assumption of a constant height of the pelvis during the double support phases did not seem to be valid for most of our subjects, an underestimation of the real vertical displacement of the pelvis during the stride cycle will be introduced by this calculation. To correct for this, we made an additional assumption about the vertical displacement during the period of double support. When it is assumed that the pelvic displacement during double support can also be approximated as an (inverted) circular path, and that no discontinuities exist in the trajectory during the transitions from single to double support (and vice versa), the total vertical displacement of the pelvis can be estimated as the sum of $\Delta\, y_1$ and $\Delta\, y_2$ (see figure 1, lower part). $\Delta\, y_2$, being the vertical displacement during double support, becomes:

$$\Delta\, y_2 = \ell'\left(1 - \cos\theta\right) \tag{3}$$

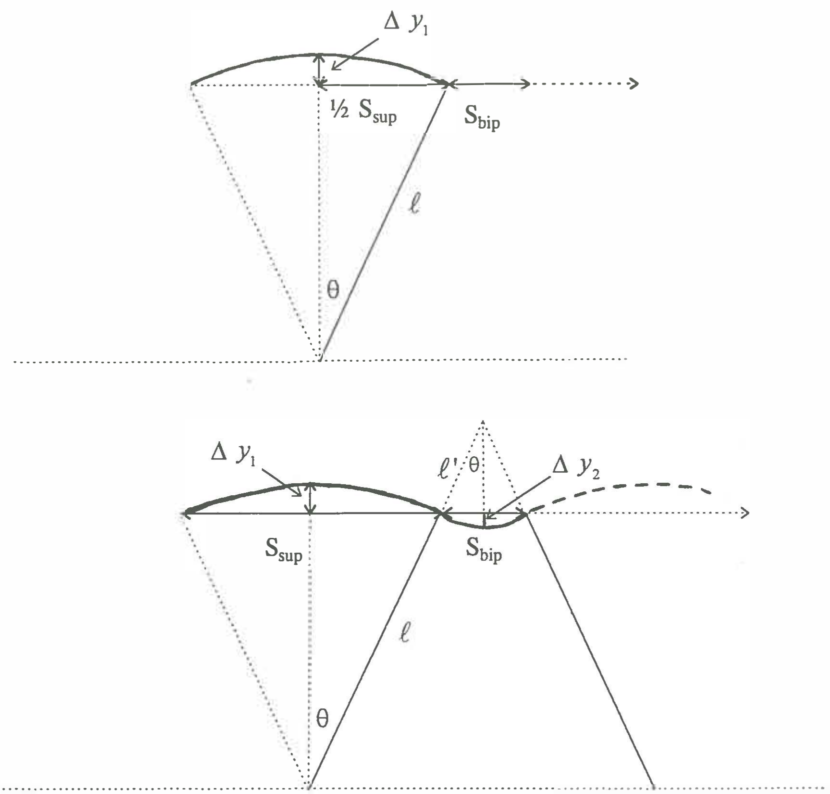

Figure 1: Models of the vertical displacement of the pelvis during walking. S_{sup} and S_{bip} represent the forward displacement of the pelvis during single and double support, respectively. The upper figure represents model I, the lower model II. The trajectory of vertical displacement of the pelvis is in both figures indicated by a bold line. The direction of progression is to the right. For further explanation, see text.

Thus, the amplitude of the fundamental intrinsic harmonic for vertical displacement (a_{y2}) can be predicted according to two models:

$$a_{y2} = 0.5\ \Delta\ y_1 \qquad \text{(model I) or,} \qquad (4)$$

$$a_{y2} = 0.5\left(\Delta\ y_1 + \Delta\ y_2\right) \qquad \text{(model II).} \qquad (5)$$

In both models θ can be calculated from:

$$\sin\theta = \frac{S_{sup}}{2\ell} \qquad (6)$$

From figure 1 (lower part), it can be seen that:

$$\frac{l'}{l} = \frac{S_{bip}}{S_{sup}} \qquad (7)$$

Thus, the only variables determining the predicted amplitude of vertical displacement are S_{sup} in model I, and S_{sup} and S_{bip} in model II. As both S_{sup} and S_{bip} are determined by stride length, it is clear that in both models the predicted vertical displacements depend on stride length only. A prediction for the phase of the fundamental intrinsic harmonic can be obtained based upon the timing of the maximal, or minimal, vertical position of the pelvis. According to both models, a maximal vertical position should be reached during the middle of single support, and a minimum should be reached during the (middle of the) double support phase.

Anterior-posterior displacement

During walking, the potential and kinetic energy of the trunk vary due to up and down movements and speed fluctuations in forward direction, respectively. In energetically optimal walking, a 100% recovery of potential and kinetic energy would lead to a constant sum of potential and kinetic energy levels throughout the stride cycle[12]. This would require that the changes in potential and kinetic energy levels during the stride cycle are exactly out of phase and equal in amplitude and shape. Although this is not entirely true during normal walking, it can be assumed, in agreement with experimental results[12,13], that the maxima of potential and kinetic energy of the trunk are approximately equal. Thus, a prediction of the fluctuations in relative anterior-posterior position of the pelvis during walking can be made as follows:

$$\Delta\left(m\,g\,y\right)=\Delta\left(\frac{1}{2}m\,v_x^2\right) \tag{8}$$

and
$$mg\,\Delta y \approx m\,\overline{v}_x\,\Delta\,v_x \tag{9}$$

$\Delta\ y$ is the vertical displacement calculated according to one of both models from the previous section (equation (4) or equation (5)). $\overline{v}_x$ is treadmill belt speed, and $\Delta\ v_x$ indicates the fluctuations in forward speed of the COM. m and g represent the subjects' mass and the acceleration due to gravity (9.81 $m.s^{-2}$), respectively. As we use the amplitude of the fundamental intrinsic harmonic (a_{x2}) to describe the relative anterior-posterior displacement of the pelvis, the displacement (x) and velocity ($\dot{x}$) functions can be deduced from equation (1):

$$4\pi\,f_c\,a_{x2}=\frac{1}{2}\Delta\,v_x \tag{10}$$

Combining this with equation (9), and substituting the product of stride frequency and stride length for mean walking speed, results in:

$$a_{x2}=\frac{g\Delta y}{8\pi\,f_c^2\,S_c} \tag{11}$$

From equation (11), it is clear that the amplitude of anterior-posterior displacement depends on both stride frequency and stride length. Assuming a 100% recovery of energy, the timing of the maximal vertical position of the pelvis should coincide with a minimal forward velocity, and the timing of the minimal vertical position should coincide with a maximal forward velocity. Consequently, the timing of the maximal forward position of the pelvis should be reached halfway the middle of single and the middle of double support.

Displacement in left-right direction

During locomotion the COM trajectory has to be controlled within narrow limits to maintain dynamic equilibrium. It has been shown that during walking the COM projection on the horizontal plane shows an approximately sinusoidal trajectory between the medial borders of the supporting feet[3,4,14]. The COM trajectory is ultimately determined by the ground reaction forces at the feet and the changes in position of body segments relative to each other. Therefore, foot placement plays a decisive role in the control of the COM trajectory during walking[15-18]. As the calculated trajectory of the point intermediate to both SIPS approximately will correspond to that of the body's centre of mass, the pelvic trajectory in the frontal plane can be estimated based upon a simple model. Assuming that the sinusoidal left-right trajectory of the COM solely is determined by foot placement and the ground reaction forces working on the foot, the relevant parameters for predicting the trajectory in left-right direction of the COM (and approximately also the pelvis) can be modelled as in figure 2. The model makes use of two basic assumptions: the centre of pressure (COP) always lies directly under the foot and stride width is approximately constant. Furthermore, it can be assumed that to ensure lateral stability the force vector F_{yz} should be directed from the COP (at the foot) towards the COM. It can then be deduced from the model in figure 2 that:

$$\frac{a_z}{g}=\frac{z_0-z}{h} \qquad (12)$$

in which, a_z represents the acceleration in left-right direction, g is the acceleration due to gravity (9.81 m.s^{-2}), h is the approximate height of the COM, z_0 is half the stride width, and z is the left-right displacement of the COM. As we expect the amplitude of the first harmonic (a_{z1}) to describe the left-right displacement of the pelvis, it can be deduced from equations (1) and (12) that:

$$\frac{4\pi^2 f_c^2 a_{z1}}{g}=\frac{z_0-a_{z1}}{h} \qquad (13)$$

Therefore a_{z1} can be estimated as follows:

$$a_{z1}=\frac{z_0}{1+4\pi^2 f_c^2 \dfrac{h}{g}} \qquad (14)$$

Equation (14) predicts that the amplitude of the sinusoidal left-right movements of the pelvis depends on stride frequency and stride width only. The equation is very similar to the equations used by Kodde et al.[19,20], for predicting the lateral movements of the COM during quiet stance on a force platform. Equation (14) has one unknown parameter; as we did not measure stride width, z_0 has to be fitted to experimental data in order to obtain optimal predictions for the amplitude of left-right movement. A prediction for the phase of the fundamental intrinsic harmonic can be obtained based upon the expected timing of the maximal left or right position of the pelvis. These maxima will be reached at approximately the middle of single support of the left and right leg, respectively.

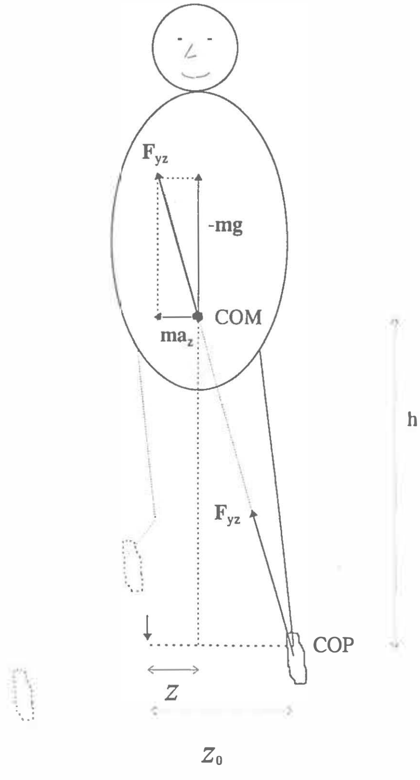

Figure 2: Model of the control of left-right movements of the COM during walking. The ground reaction forces acting on the foot are indicated by the force vector F_{yz}. Its direction is from the COP at the foot towards the COM. a_z represents the acceleration in left-right direction, g is acceleration due to gravity (9.81 $m.s^{-2}$), m is the subject's mass, h is the height of the COM, z_0 is half the stridewidth, and z is the left-right displacement of the COM. The figure depicts the lateral movement relative to the midline of locomotion (indicated by an arrow), in a subsequent step (indicated by the footprint) a similar lateral movement will be made in contra-lateral direction.

Results

Adaptation of spatio-temporal parameters of the stride cycle

Although the amount of adaptation of stride length and frequency clearly differed between individuals, none of our subjects experienced any difficulties in performing this task. As illustrated by figure 3, the mean voluntary increase or decrease in stride frequency (and vice versa stride length) at a particular speed was similar over all treadmill belt speeds. The adaptation of the walking pattern was largest when step length had to be decreased. The mean adaptation of stride frequency was +0.12 Hz in the conditions were the subjects had to make small steps and -0.10 Hz in the large steps conditions. Only marginal differences in the amount of adaptation were observed between conditions with different speeds.

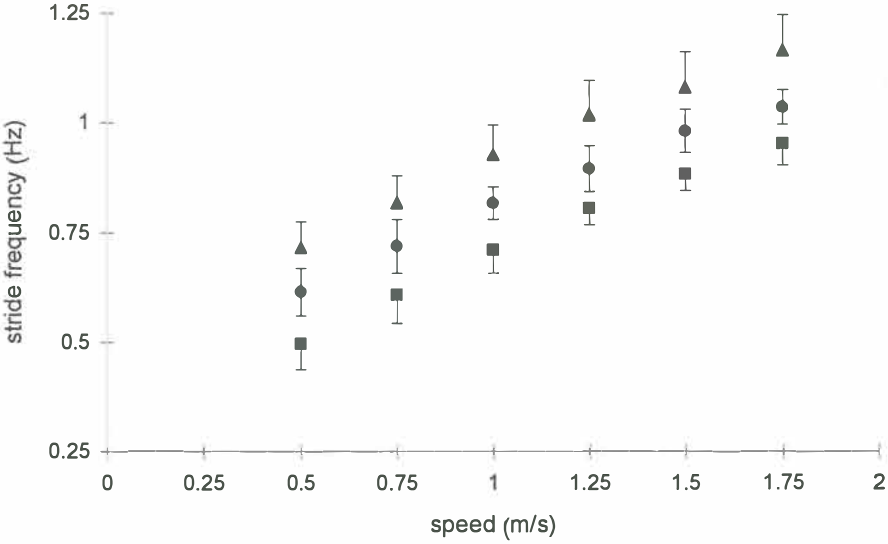

Figure 3: Mean stride frequencies during normal treadmill walking and in conditions where step length had to be modulated. The different symbols indicate group means (n=10) during normal walking (•), walking with large steps (■) and walking with small steps (▲). Standard deviations are indicated by a bar.

With changes in speed, stride frequency, and stride length, variations in the durations of support, swing and double support phases could be observed. The latter changes resulted in changes in the relative timing of the stride cycle which depended on stride length. In all subjects and regardless of the condition, increases in stride length resulted in a decreasing relative duration of the support phase, and an increase in the relative duration of the swing phase. For the group mean values, the relation between stride length (S_c) and the relative duration of the support phase ($T_{sup\%}$) could be described by a linear regression function: $T_{sup\%}$= -6.61*S_c+76.6 (R^2=0.90, $p<0.001$).

The pattern of displacement of the pelvis during walking

A typical example of the individual data obtained during treadmill walking is presented in figure 4. The dashed vertical lines in figure 4 indicate the start and end of a stride cycle. The patterns of displacement between these lines represent examples of the displacement curves which were described by means of the harmonic analyses. The figure clearly indicates the oscillatory nature of pelvic displacement during walking, but also slight deviations from a perfect periodicity and symmetry of pelvic displacement can be seen. Small fluctuations in the pattern of displacement from stride cycle to stride cycle will be reflected in the calculated coefficients of variation for the harmonics. Any asymmetry of pelvic displacement will manifest itself in a contribution of extrinsic harmonics in the description of the displacement curve. The degree of correspondence of displacement curves to the hypothetical perfect periodicity and symmetry of pelvic movements during walking will be elaborated in the next sections.

The harmonic analyses: contributions of intrinsic and extrinsic harmonics

Before giving a full presentation of the variations in amplitudes of intrinsic and extrinsic harmonics in the experimental conditions, we will present a general report of the results of the harmonic analyses in this section. The calculated x, y and z periodic displacement functions yielded good descriptions of the actual displacement curves with only a limited amount of harmonics. We generally found that the first five harmonics (or less) sufficed for describing at least 99% of the variance of the original displacement curve. The fundamental intrinsic harmonics, i.e. the first harmonic for left-right displacement and the second harmonics for vertical and anterior-posterior displacements, always were largest, and always showed the least cycle-to-cycle variability. As will become clear in the subsequent sections, the contributions of higher intrinsic harmonics mostly served to modify the shape of the calculated displacement curves, they did not substantially contribute to the amplitude of displacement. Their variability generally increased with increasing harmonic number. Extrinsic harmonics (of lower order) appeared to play a role only in the vertical and anterior-posterior displacements. When the amplitudes of the fundamental intrinsic harmonics were compared to the displacement amplitudes as calculated directly from the displacement curves, all trends were similar, and no differences larger than 5 mm were observed.

Figure 5 is a summary of the results of the harmonic analyses of the displacement curves, it presents the mean amplitudes and the timing of the first maximum of the fundamental intrinsic harmonics (x2, y2, and z1) in all conditions. Systematic influences of conditions are obvious in all fundamental harmonics and in both amplitude and timing. The influence of deviations of the preferred walking pattern and treadmill belt speed upon the amplitude of fundamental harmonics can be deduced from figure 5, but will be discussed further in the next three sections. The timing of maxima of the fundamental harmonics shows fairly consistent values for all three movement directions. As indicated by figures 4 and 5, maximal left-right displacement and the highest vertical position are reached during single support, in both cases approximately at the middle of the single support phase. Maximal forward displacement is reached at the end of double support or during the transition from double to single support (indicated by (-) in figure 5). This is a consistent departure from the model predictions indicated by (×). Although small effects of adaptation of stride frequency (and stride length) upon the phase of the fundamental harmonics can be seen in all movement directions, these effects can largely be attributed to the changes in relative timing of the stride cycle (as indicated in figure 5). Only for the displacement in left-right direction are there some small deviations from the timing as predicted in the figure.

Displacement in vertical direction

The fundamental intrinsic harmonics of the displacement functions for vertical directions clearly accounted for the largest part of the variance in the actual displacement curves. Regardless of the frequency/amplitude combination, the fundamental harmonic on average accounted for 70-75% of the variance in vertical position at the lowest speed (0.5 m/s). This percentage increased to 86-89% at a treadmill belt speed of 0.75 m/s, and further increased with increasing speed to approximately 95-98% at the four highest treadmill belt speeds. The lower percentages at the two lowest speeds were largely due to a relatively high contribution of the first harmonic at these speeds. The contribution of this extrinsic harmonic reflects an asymmetry of the vertical pelvic trajectory which was present in all subjects. However, in 8 out of 10 subjects the first harmonic only made a minor contribution to the displacement function at speeds higher than 1 m/s. Adding the contributions of the first, second and fourth harmonics yielded descriptions which always accounted for 98% or more of the variance. Thus, the third and fifth harmonics only marginally contributed to the description of the displacement curves.

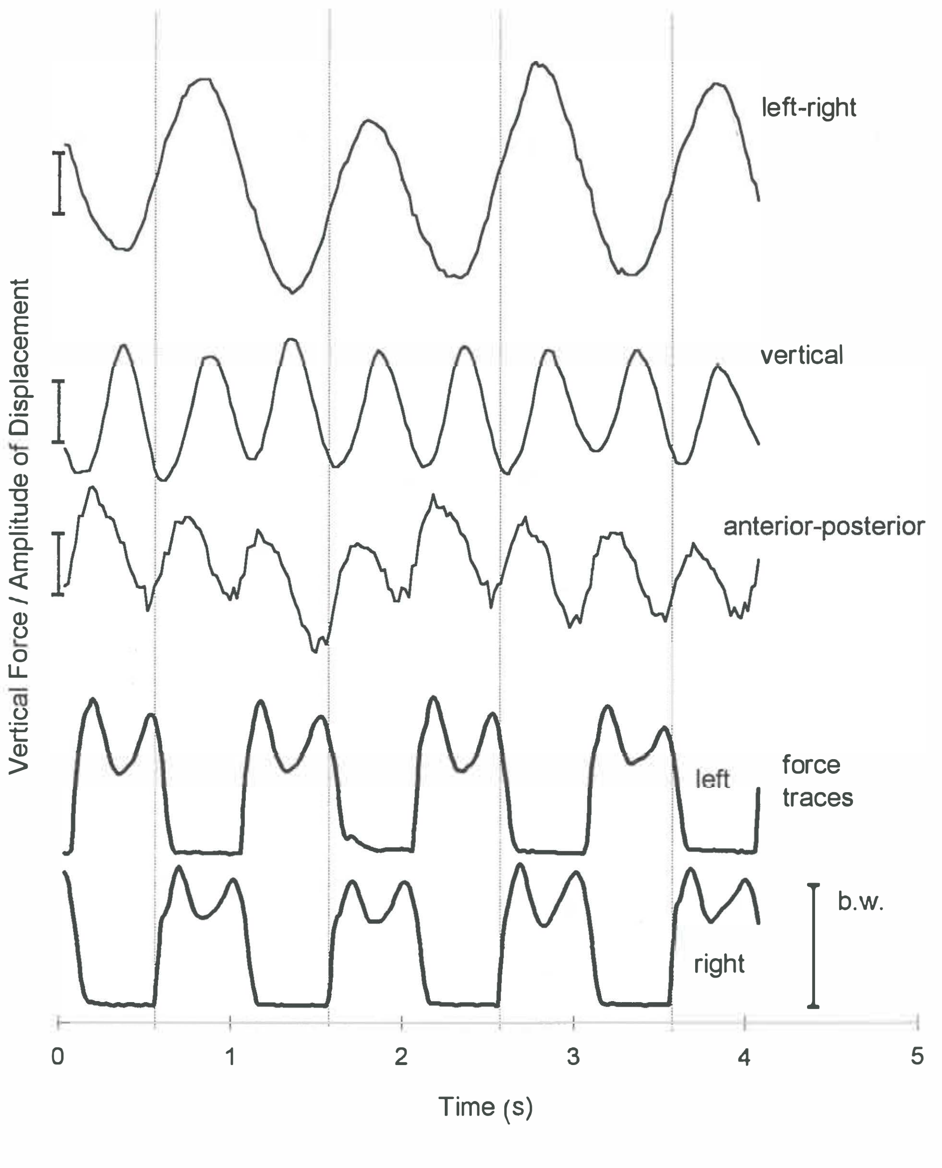

Figure 4: Representative example of data obtained during treadmill walking. The subject walked with a spontaneously chosen combination of stride length and frequency at a treadmill belt speed of 1.5 m/s. The upper three traces indicate displacements of the pelvis, the lower two traces indicate left and right vertical force components. The dashed vertical lines in the figure indicate the start and end of a stride cycle. In the upper left part of the figure, the three bars each indicate 20 mm. In the lower right part of the figure, a bar indicates body weight (b.w.).

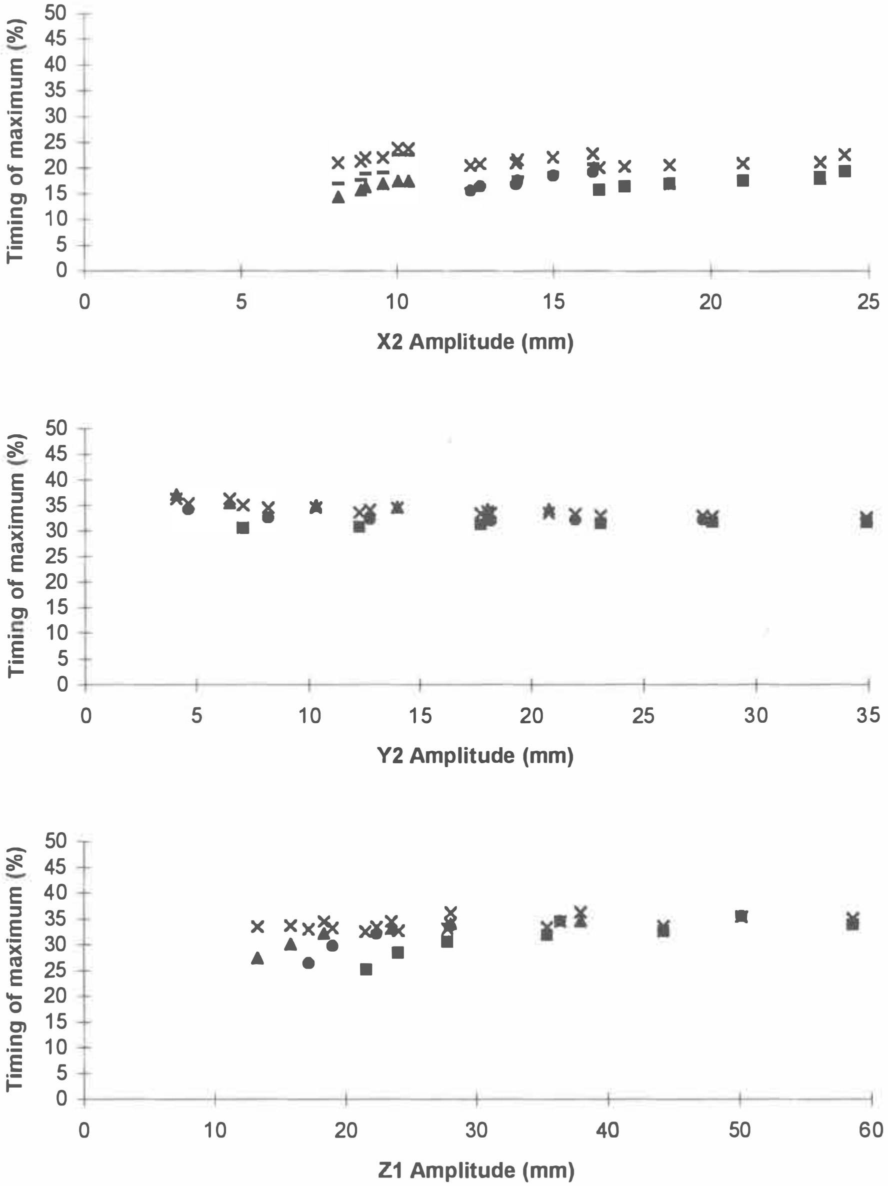

Figure 5: Amplitude and timing of maximal displacement of the pelvis during treadmill walking in different conditions. The different symbols indicate group mean values for the timing and amplitude of the fundamental intrinsic harmonics (x2, y2, and z1) during normal walking (•), walking with large steps (■) and walking with small steps (▲). (×) indicates the timing of the mid of single support in the two lower figures, and an instant halfway the middle of single and the middle of double support in the upper figure. The upper figure also indicates the transition from double to single support (-). The vertical axis indicates the timing of the stride cycle starting with a right support phase (0%) and ending with the start of the left support phase (50%). The upper figure represents the timing and amplitude of the maximal forward position of the pelvis (as indicated by the fundamental harmonic), the middle figure indicates the highest vertical position, and the lowest figure indicates the maximal displacement to the right.

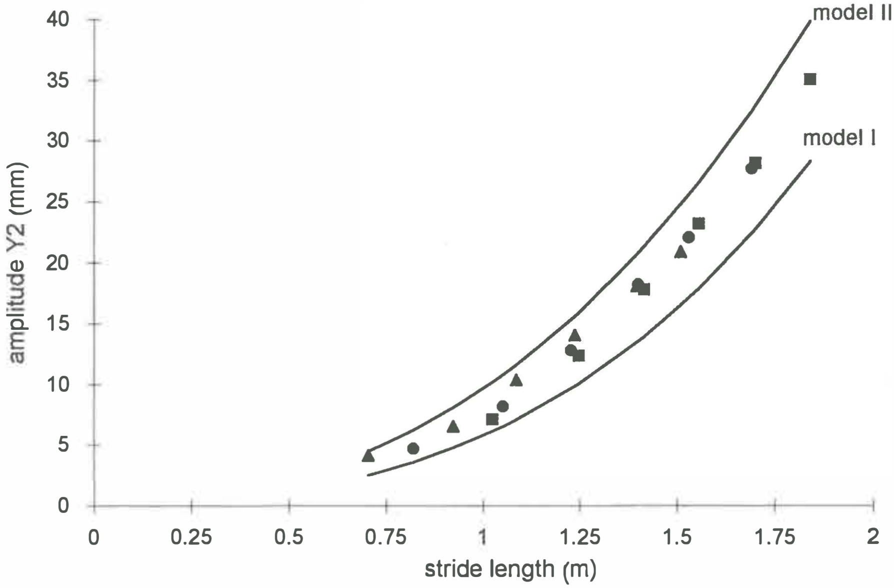

Figure 6: Mean amplitudes of the fundamental intrinsic harmonic for vertical displacement (y2) in the different conditions. The amplitudes predicted according to two different models of vertical displacement during walking are represented as two black lines. Symbols indicate group means (n=10) during normal walking (•), walking with large steps (■) and walking with small steps (▲).

Figure 6 shows mean amplitudes of the fundamental intrinsic harmonic in the different conditions and, as two black lines, the predicted amplitudes of the vertical displacement according to the two different models discussed in a previous methods section. To obtain smooth lines indicating the predicted values, a regression equation linking the group mean values for S_{sup} and S_{bip} to stride length was used rather than the actual measured values for these variables. As predicted, a stride length dependency of the amplitude of vertical movement of the pelvis is seen. As already foreseen, the first model clearly underestimates the actual values. However, the second model (which additionally takes into account a downward displacement of the pelvis during the double support phases) tends to overestimate the actual values. In fact, it was seen in the individual data that three subjects behaved more according to model I, while four other subjects showed better correspondence to the values predicted by model II. The remaining three subjects showed an intermediate type of behaviour. That the mean values for all subjects lie between the predictions of both models is therefore not surprising. The predictive value of both models was tested by means of linear regression equations of the observed mean amplitudes upon the predicted values. The individual predictions generated R^2 values ranging from 0.94 to 0.99 (for the subjects which behaved according to model I) and 0.97 to 0.99 (for the subjects which behaved according to model II). The remaining subjects showed R^2 values which ranged from 0.79 to 0.99.

Only a small contribution of the fourth harmonic was seen; its amplitude varied between 0.6 and 2.2 mm. The amplitudes of the first harmonic were low also, but they increased with stride length from 1.7 to 3.8 mm. The third harmonic also increased with stride length, but the amplitudes of this harmonic did not exceed 1 mm. Amplitudes of the fifth harmonic were always lower than 0.5 mm. The (relatively) large amplitudes of the first (and third) harmonic at low speeds, together with the strong increase of the second harmonic (see figure 6) and marginal increases in the first (and third) harmonic amplitudes with stride length, explain the increasing percentage of variance accounted for by the fundamental intrinsic harmonic.

The cycle-to-cycle variability in the amplitudes of the first five harmonics was analysed by means of the calculated coefficients of variation. These coefficients were only moderately low for the fundamental intrinsic harmonic: for y2 the coefficients of variation were highest at the lowest speed (between 30-33% at 0.5 m/s), and remained approximately constant within a range of 10-20% at all higher speeds. At these speeds, the high frequency (small steps) conditions consistently had a higher variability. Within the range already mentioned, the coefficients of variation in these conditions had values which were at least 5% higher than the values for normal walking and large steps conditions. The latter two conditions always had very similar values. The coefficients of variation for the fourth harmonic exceeded 50% in all conditions, values for the extrinsic harmonics (y1, y3, y5) were excessively large; they always were larger than 100%.

Displacement in anterior-posterior direction

The mean percentage of variance in anterior-posterior position accounted for by the fundamental intrinsic harmonic (x2) fluctuated between 86-94% and did not show an influence of treadmill belt speed. However, within this range, the displacement curves in conditions where the subjects had to make small steps were less well (approximately 5% less) accounted for by the fundamental intrinsic harmonic. Many subjects showed a contribution of the first (extrinsic) harmonic over the whole range of speeds, the mean values of the contribution of this harmonic ranged between 4-12% over all conditions. In all conditions, a small contribution of the fourth harmonic was seen. The magnitude of its contribution varied between 1-4%, and tended to increase with speed. As the third and fifth harmonics hardly contributed to the descriptions of the displacement curves, an accurate description was obtained by the sums of the first, the second and the fourth harmonics .The mean percentage of variance explained by these harmonics was minimally 99%.

Figure 7 shows the mean amplitudes of the fundamental intrinsic harmonic (x2) as a function of treadmill belt speed. Different conditions are indicated by the different symbols. Closed symbols represent empirical data, open symbols represent the predicted amplitudes. For the model predictions shown in this figure, the predicted vertical displacements according to model II which included vertical displacements during both single and double support were used. Although the model predictions tend to overestimate the actual values in the conditions where subjects had to make large steps, the observed trends in empirical data are well predicted by our model. Determination of the predictive value of the model this time yielded an R^2 value of 0.97 for the group means. For the individual predictions, the R^2 values varied between 0.88 and 0.97.

Particularly when subjects made large steps at lower speeds, the first harmonic significantly contributed to the displacement functions. At the two lowest speeds the amplitude of this harmonic varied between 2.9 and 8.1 mm. At higher speeds, values never exceeded 4.2 mm, and minimal values were obtained in conditions where subjects walked normally or made small steps (for the latter conditions the minimal value was 2.2 mm). The fourth harmonic slightly increased with speed in all conditions. Overall, its amplitude varied between 1.1 and 3.2 mm. Within this range, consistent differences were seen between conditions: the amplitudes of the fourth harmonic were lowest in the small steps conditions, highest in large steps conditions and always intermediate for normal conditions. Contributions of the third and fifth harmonics can be neglected, their amplitudes never exceeded 0.75 mm.

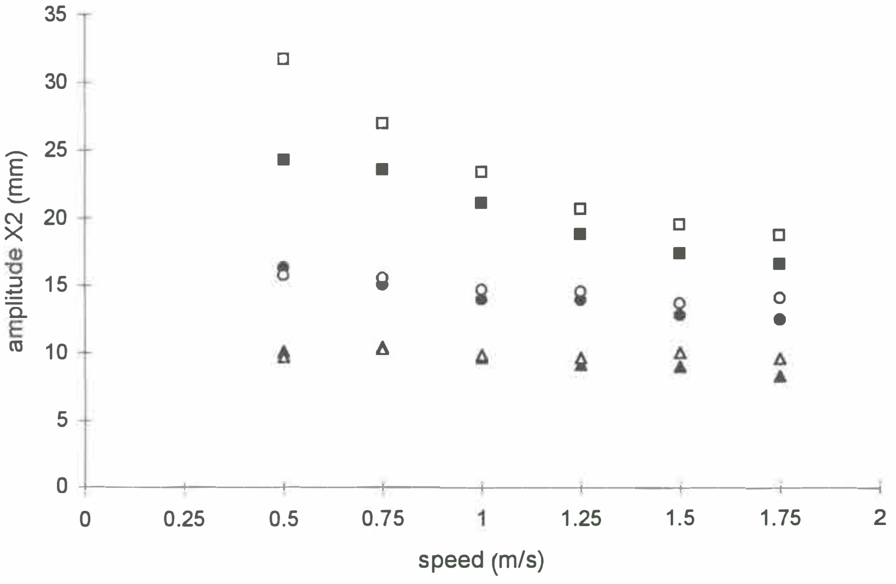

Figure 7: Mean amplitudes of the fundamental intrinsic harmonic for anterior-posterior displacement (x2) as a function of treadmill belt speed. Different conditions are indicated by the different symbols. Symbols indicate group means (n=10) during normal walking (•), walking with large steps (■) and walking with small steps (▲). Closed symbols represent empirical data, open symbols represent the predicted amplitudes.

Coefficients of variation of the fundamental intrinsic harmonic (x2) were always lowest during normal walking, and during the conditions where subjects made large steps. At treadmill belt speeds above 0.5 m/s, the values for these conditions remained constant within a range of 12-14.5%. Within the same speed range the high frequency (small steps) conditions had a variability between 17.5 and 22.5%. At a treadmill belt speed of 0.5 m/s, the variability coefficients were higher for all conditions (16.9, 24.3 and 25.2 % for normal walking, large and small step conditions, respectively). Coefficients of variation for the fourth harmonic were larger than 50%, and the values for the extrinsic harmonics were again excessively large (more than 150%).

Displacement in left-right direction

On average, extrinsic harmonics did not play any role of importance in the displacement functions for left-right movements. In all conditions, the displacement curves in left-right direction were almost totally accounted for by the fundamental intrinsic harmonic. At the four lowest treadmill belt speeds, this harmonic, on average, accounted for at least 97% of the variance of the displacement curves. At the two highest speeds, this percentage decreased somewhat, but it remained higher than 90%. When the third harmonic was taken in addition to the fundamental intrinsic harmonic, the percentages accounted variance increased to a minimum level of 98%. In the few conditions where the first and third harmonics explained "only" 98%, the fifth harmonic often added the last percents.

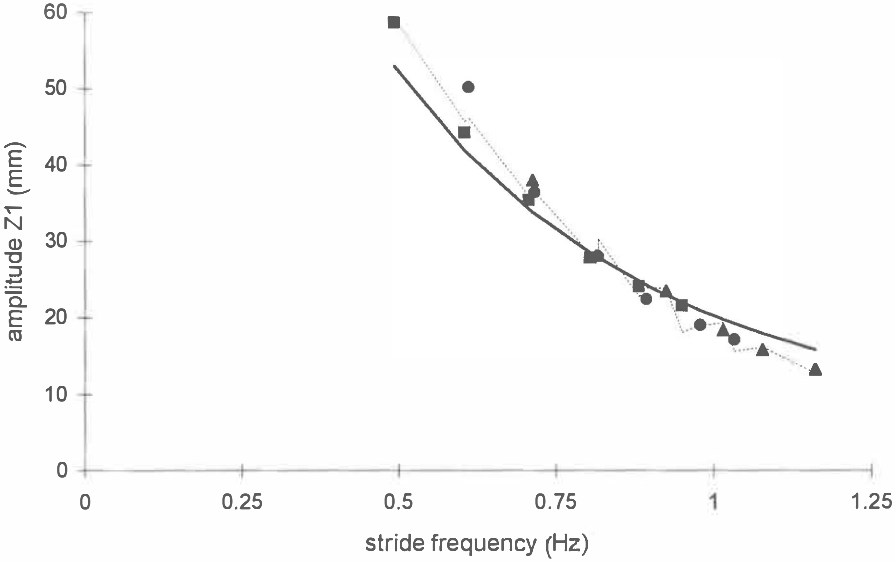

Figure 8: Mean amplitudes of the fundamental intrinsic harmonic for left-right displacement (z1) in the different conditions. The predicted amplitudes of left-right displacement are shown as a black line. Symbols indicate group means (n=10) during normal walking (•), walking with large steps (■) and walking with small steps (▲). The dashed line indicates an improved model prediction to which is referred to in the discussion.

Figure 8 shows mean amplitudes of the fundamental intrinsic harmonic (z1) in the different conditions and as a black line, the predicted amplitudes of left-right displacement as predicted for the group mean stride frequencies in different conditions. The predictions were made assuming a constant stride width of 0.22 m and a mean pelvic height of 1.10 m. A confirmation of the predicted dependency of left-right pelvic movements of stride frequency is evident. The validity of the model predictions made for the individual and group mean values was tested by means of a linear regression of the observed mean amplitudes upon the predicted values. For the group means this yielded an R^2 value of 0.98, for the individual predictions the R^2 values ranged between 0.88 and 0.99. Consistent between subject differences were observed in stride width values which yielded the best predictions of left-right movement in an individual. The stride widths used for the individual predictions varied between 0.14 and 0.28 m.

The amplitudes of harmonics other than the fundamental harmonic were always small. Amplitudes of the third and fifth harmonic varied between 1.1 to 3.6 and 0.4 to 1.7 mm, respectively. Within these ranges, the amplitudes of both third and fifth harmonic tended to increase with stride frequency. This explains the earlier mentioned decrease in the percentage of variance accounted for by the fundamental harmonic. Amplitudes of the extrinsic harmonics (z2, z4) were very small: the second harmonic varied between 0.6 to 1.6 mm and the fourth harmonic was always lower than 0.5 mm.

Only for the fundamental intrinsic harmonic were the coefficients of variation moderately low: for z1 the coefficients of variation increased with speed from approximately 10-15% at 0.5 m/s to 20-30% at 1.75 m/s. The high frequency (small steps) conditions most often were characterised by a somewhat higher variability. The coefficients of variation for z3 and z5 generally exceeded 50%, and they were excessively large for z2 and z4 (always larger than 100%).

Discussion

Our analysis of the 3D displacements of the pelvis clearly shows that the displacements of the pelvis can to a large extent be described by the fundamental intrinsic harmonics. Whether subjects walked with spontaneously chosen or voluntarily adapted combinations of stride length and frequency, the fundamental intrinsic harmonics always accounted for most of the actual displacement of the pelvis. This was true regardless of the direction of pelvic displacement. Considering this, and considering the absence of large differences with actual displacements, the fundamental intrinsic harmonic seems to be an adequate description of at least the amplitude of pelvic displacements. Nevertheless, in all movement directions a contribution of higher harmonics was seen when it was attempted to adequately describe the shape of the oscillatory movements. However, an adequate description did not require more than the first five harmonics. A relevant contribution of extrinsic harmonics to the description of pelvic displacement was only observed in the vertical and anterior-posterior displacement of the pelvis, particularly at the lower speeds. The magnitude of this contribution not only varied strongly between individuals, but as indicated by the coefficients of variation, also from stride cycle to stride cycle. The contribution of extrinsic harmonics to the description of anterior-posterior and vertical displacements is in agreement with the notion that extrinsic harmonics reflect asymmetries due to differences in left and right step duration or length, as was also suggested by Capozzo[5].

For the comparison of model predictions with actual displacement of the pelvis we used the amplitude of the fundamental intrinsic harmonic as a measure of the actual displacement. The results of our harmonic analysis, and particularly the comparison of the amplitude of the fundamental intrinsic harmonic with the calculated displacement amplitudes underline the validity of this approach. Further support for the validity of taking the amplitude of the fundamental intrinsic harmonic as a measure of the actual amplitude of pelvic oscillations is seen when the fundamental intrinsic harmonic amplitudes presented in our results are compared with the amplitudes of left-right, vertical, and anterior-posterior displacement of the pelvis[7], or displacement data of the lower spine (L3)[6] during treadmill walking. Our results on normal treadmill walking and the results of these studies generally show good agreement. There is one major exception; increases in the amplitude of left-right movements were reported[6,7] during walking at higher treadmill belt speeds than those in our study. The latter increase in the amplitude of left-right movements of the pelvis with treadmill belt speed can not be predicted from the empirical and model data presented here. The increase in left-right movements at treadmill belt speeds beyond the speed where usually the transition from walking to running is observed suggests that our model predictions may only be valid within a limited range of walking speeds and that additional mechanisms play a role at higher speeds. For normal adult men, a transition from treadmill walking to running usually takes place at a speed of about 1.88 m/s (Ref. 21). Thus, the speed range in our study (0.50 to 1.75 m/s) at least includes the full range of normal walking speeds.

A good correspondence is seen between our data on normal treadmill walking and those of Capozzo[5] who studied displacements of the head and trunk during free walking at different speeds by means of the same methodology as we used in our study. When Capozzo's and our results are compared within a similar speed range, the changes in amplitudes of the fundamental harmonic with walking (or treadmill belt) speed are very alike for all directions of displacement. Only for the vertical displacement are some differences seen, the fundamental intrinsic harmonic amplitudes as reported by Capozzo are consistently higher than ours. A likely explanation for this difference is the somewhat higher step length which has been reported for free walking in comparison to treadmill walking at similar speeds[22], and possibly also the fact that Capozzo reports individual data on five subjects (and few stride cycles) only.

Thus, despite this difference with Capozzo's data on free walking, it seems reasonable to assume that the presented kinematics of treadmill walking do not differ from those who would have been obtained during free walking with similar stride lengths and frequencies. Although the biomechanical similarity between treadmill and overground walking is also emphasised by other authors[23], small differences between kinematic data on treadmill walking and overground walking are possible as a result of speed fluctuations of the treadmill[23,24].

Whereas earlier studies only described the displacements of the pelvis (or HAT) as a function of walking speed, our results unequivocally demonstrate that the pattern of displacements of the pelvis is predominantly determined by either stride frequency (left-right movements) or stride length (movements along the vertical axis). The models used for predicting left-right and vertical movements of the pelvis generally give good predictions of the amplitudes of the oscillatory pelvic movements. Displacements in anterior-posterior direction were also predicted reasonably well by a model. They are dependent on both stride frequency and stride length.

Despite the generally good validity of the presented models, a number of shortcomings are apparent. First of all, we are aware that, although COM and pelvic displacement may approximately correspond to sinusoidal movement patterns, the underlying pattern of linear acceleration will certainly not be perfectly described by a sinusoidal function (e.g. Ref. 4). However, in experiments where we used accelerometers on the pelvis, we were able to calculate similar displacement patterns as described in the present study, despite the fact that the frequency content of the measured acceleration signals was much higher than that of the displacement signal.

Secondly, in the model for left-right displacement, the simplifying assumption of a constant vertical force vector equal, but opposite in direction, to the acceleration due to gravity also needs discussion. An analysis of the ground reaction forces acting on the feet shows that the real vertical force component changes during the stride cycle, and that the interaction between horizontal and vertical forces changes with speed[25,26]. We therefore tried to improve the model predictions by making the additional assumption that changes in the vertical force vector with speed can be estimated by subtracting the centripetal forces from the gravitational forces working on the body. The centripetal acceleration is due to the circular trajectory of the body over the supporting leg, and it is equal to v^2 / ℓ . Thus, when g in equation (14) is exchanged for $(g - \bar{v}_x^2 / \ell)$, new predictions can be made as indicated by the dashed line in figure 8. It can be seen that this model correction indeed leads to a greater accuracy in predicting the amplitude of the fundamental harmonic for left-right displacement, albeit at the cost of some simplicity.

Thirdly, by means of the models, we have made predictions for the mean values of displacement during a number of stride cycles. For the predictions of left-right displacement, we had to assume a constant stride width whereas from stride cycle to stride cycle and, perhaps even more so, from condition to condition stride width may have varied. It is likely that a large part of the variance in left-right movements, as expressed by the coefficients of variation of the fundamental harmonic, reflects variations due to a cycle-to-cycle control of the COM trajectory by means of small variations in foot placement. In our analysis of pelvic displacements during walking we did not specifically pay attention to non-periodic changes in the movement pattern which evolved over a number of stride cycles. This is partly due to the fact that we did not measure certain aspects of the movement pattern (like, for instance, foot placement), but this must also be attributed to the chosen methodology; by means of our harmonic analyses we focused on the periodicity of the kinematic patterns.

In our prediction of the *timing* of anterior-posterior displacements we made the simplifying assumption of a constancy in the mechanical energy of the COM. As the possibility of speed fluctuations of the treadmill can not be ruled out[26], the kinetic energy of the COM may have shown small deviations from the kinetic energy which can be calculated when a constant treadmill belt speed is assumed. Not only this factor would introduce a small error, during walking the COM does show slight variations in mechanical energy[12,13]. It has been shown that the peak forward velocity is attained at, approximately, heel contact[13]. This means that the phase of the changes in velocity is slightly advanced compared to what is predicted when a 100% recovery of mechanical energy is assumed. The maximal forward position would therefore also be reached earlier than predicted. This probably explains the small error in the predicted timing of maxima in figure 5. Experimentally, we observed that the transition from double to single support was a better predictor for the timing of maximal forward displacement (see also figure 5).

In conclusion: In this study the normal pattern of pelvic displacements has been described for a large range of walking speeds and a large number of different combinations of stride length and stride frequency. The results of the harmonic analysis of the displacement curves per stride cycle not only confirm the sinusoidal character of the pelvic trajectory during walking, they also extend existing data of pelvic kinematics to locomotion modes which deviate from the normal preferred walking pattern. The presented data define the normal movement behaviour of the pelvis during walking, and, thus, may be useful as normative data for situations where normal and pathological locomotion are to be compared. The presented models give insight into the underlying mechanical mechanisms which determine the pelvic trajectory. As the head, arms, and trunk are all supported by the pelvis, the pelvic trajectory determines to a large extent the effect any movement of head, trunk or arm segments may have on the COM trajectory. Therefore, knowledge of the pelvic kinematics during walking is essential in understanding the regular locomotor pattern and the mechanisms which play a role in equilibrium control during walking.

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Chapter 7: General Discussion

Introduction

The studies presented in this thesis have explored characteristics of human walking in dependence of walking speed, stride length and stride frequency. In this general discussion, the results of these studies will be evaluated in regard to the purposes mentioned in the general introduction of this thesis. A first section will regard variant and invariant aspects of human stepping patterns. This section summarises the observed relationships between spatio-temporal parameters of the stride cycle in different conditions, and the observed effects of amplitude and frequency of stepping movements on other gait parameters. A subsequent section is directed at voluntary and involuntary mechanisms which play a role in the adaptability of gait. This section will discuss the extent to which the characteristics of the observed stepping patterns are determined by central or mechanical factors. Hereafter, gait disturbances in Parkinson's diseases will be shortly discussed. Subsequent sections will discuss the results presented in this thesis in regard to clinical gait analysis.

Variant and invariant aspects of human stepping patterns

As illustrated in chapter 1, studying variant and invariant aspects of movements executed in different conditions may yield information about the coordinative principles involved. This section will consider variant and invariant aspects of spatio-temporal gait parameters in the different conditions studied in this thesis. Chapters 2 and 3 confirm that during normal walking, increases in speed are achieved by increasing both frequency and amplitude of stepping movements. Both studies show that in healthy adult subjects the relationships between measures of amplitude and frequency (and consequently also walking speed) generally have linear characteristics. This linearity is demonstrated for different groups of adult subjects, but individual results yield better descriptions by linear functions (see chapter 3 (study II)). The slightly inferior descriptions of the results for groups of subjects can be attributed to between subject differences in the range over which the three basic parameters of walking are adapted, and differences in the coefficients (i.e. slope and intercept) which define the individual linear relationships between these parameters. Provided that measurements are repeated under similar conditions, the reliability of the linear relationships is high in individuals (see chapter 2). Chapter 3 (study II) shows that treadmill walking induces consistent changes in the linear relationships. During treadmill walking, the amplitude of stepping movements at any particular speed is consistently lower than during overground walking at the same speed. However, within the range of speeds studied, linearity is preserved. Linear relationships are also observed in age-groups of children between four and ten years old (see chapter 5). This chapter demonstrates that during treadmill walking, the children generally adapt their leg movements in a way similar to adults. By means of dimensional analysis, it is shown that differences in mean spatio-temporal data between children (older than four years old) and adults can be fully attributed to differences in body size. However, individual children (particularly those in younger age groups) do not always show a linearity in spatio-temporal relationships equal to that in adults, and the intra-individual variability of temporal parameters of the stride cycle is higher in children. The former contributes to a between subject variability in children which exceeds the variability observed in adults.

Chapters 2 and 3 (study I) demonstrate condition specific effects when adult subjects adapt their stepping patterns in accordance with instructions concerning the use of external cues during stepping. In all subjects, immediate effects are observed of adapting the walking pattern to auditory cues (i.e. rhythmic walking), and stepping over visual cues (i.e. walking with visually guided step length). During rhythmic walking, step length remains almost constant whereas step frequency (and, thus, also walking speed) are increased in subsequent measurements. The relationship between step frequency and speed nevertheless can still be described well with linear functions. Walking with visually guided step length also induces different walking patterns. When step length is increased in accordance with instructions, step frequency remains constant or decreases. Again, the relationship between amplitude and speed in healthy subjects can be described well with linear functions. The specific changes in amplitude-frequency relationships during rhythmic walking and walking with visually guided step length reveal that the relative durations of support, swing, and double support phases in the stride cycle depend on amplitude of leg movements only. When step length is constant and step frequency increases (as in rhythmic walking), the relative timing of phases in the stride cycle remains constant.

Chapter 4 confirms that both stride frequency and support length (a measure of amplitude of leg movements) increase with increasing belt speed during normal treadmill walking conditions. However, in split-belt walking conditions, stride frequency is intermediate to the stride frequencies observed during normal treadmill walking at speeds corresponding to the left and right belt speeds, and the amplitude of leg movements differs between the left and right leg. As indicated by the relationships between support length and belt speed, the amplitude of leg movements during split-belt walking tends to maintain a similar relationship with belt speed as during normal treadmill walking. However, also support length is significantly influenced by contra-lateral belt speed. Thus, no invariant relationship exists between support length and belt speed. Furthermore, during split-belt walking, the relative timing of support, swing, and double support phases shows no invariances either. The only observed invariance during split-belt walking is the maintenance of an alternating pattern of leg movements with recurring periods of double support. This results from a symmetric adaptation of both double support phases. Chapter 5 demonstrates that the adaptation to split-belt walking conditions in children between four and ten years old is similar to that in adult subjects. However, the adaptation of spatio-temporal parameters is less consistent in the youngest children and, similar to normal treadmill walking conditions, both a higher intra- and inter-subject variability are noted in children.

In chapter 6, a dependency of spatio-temporal parameters of pelvic displacement from either stride length, stride frequency, or both is shown. Due to the mechanics of human walking, linear displacement of the pelvis during walking is of an oscillatory nature. Over a wide range of amplitudes, frequencies, and walking speeds, the linear displacements in the three movement directions can be described reasonably well with sinusoidal functions. The amplitude and timing of the oscillatory movements can be predicted by relatively simple mechanical models. Vertical amplitude of the oscillations is determined by stride length only. Stride frequency and stride width determine the left-right amplitude of pelvic displacement. Amplitude of anterior-posterior oscillations depends on both stride length and frequency. In all movement directions, the timing of maximum displacements is to a large part determined by the timing of support phases in the stride cycle.

From the results presented in chapters 2 to 6, it can be concluded that, within the ranges of amplitude and frequency studied, these parameters tend to scale linearly towards each other in all conditions studied. Thus, linear relationships exist also with walking speed. During normal walking conditions, rhythmic walking, walking with visual guide, and treadmill walking, the relative timing of support and swing phases linearly depends on the amplitude of leg movements. Due to an asymmetry in leg movements, this dependency disappears in split-belt walking conditions. Even though stereotypical relationships between spatio-temporal parameters of the stride cycle can be observed during normal walking, the results of the studies in other conditions demonstrate that these relationships are not invariant. The only invariance observed in all conditions is an alternating pattern of leg movements with recurring periods of double support. However, considering the fact that humans are able to switch to running, the maintenance of double support phases can not be regarded as an invariance in human locomotion. Chapter 4 shows that, even in unusual conditions like split-belt walking, human locomotion is characterised by an alternating pattern of leg movements with phase lags between the mid of left and right support phases of approximately 0.50. An analysis of the adaptation of spatio-temporal gait parameters during the first 45 strides after switching from normal treadmill walking to split-belt walking demonstrates that these phase lags remain close to 0.50 from the first stride cycle and onward[1]. All other analysed parameters, including periods of double support, settle in stable values after approximately 12 to 15 stride cycles. Yet, although adult human locomotion seems to be invariantly characterised by alternating leg movements with phase lags of approximately 0.50, these phase lags can not be regarded as invariant either. Humans are able to locomote with non-alternating patterns of movement, for example by crawling on all fours, hopping, and performing learned movement patterns like hop-step-jump (i.e. triple jump). The latter movement patterns all have phase lags other than 0.50. Therefore, these observations, in combination with the results presented in this thesis, give evidence that no invariances exist in spatio-temporal parameters of the stride cycle during human locomotion.

Voluntary and involuntary control mechanisms during human walking

During human locomotion, volition normally is directed at overall goals, for example the time when we have to be somewhere, the route we take to get there, and the walking speed necessary to arrive in time. In doing so, we normally do not think of how we do so. Yet, we can also direct our attention to almost any parameter of our movement pattern, and within certain limits (which are hitherto largely unknown) we can also voluntarily control aspects of our walking patterns beyond speed, stride length and frequency. The studies presented in thesis show that when the nature of voluntary control changes, the task itself, and also the coordinative mechanisms involved, change. This is not merely a trivial statement; the consequences of changes in voluntary control can be unexpected involuntary changes in coordination as witnessed by the studies presented in chapters 2 and 3 (study I). Hence, changes in voluntary control may reveal properties of coordinative mechanisms which remain out-of-sight during stereotypical walking conditions. Thus, volition can direct, and affect, our movements. Nevertheless, typical features of human locomotion normally all are achieved by involuntary mechanisms which do not require conscious effort. The way in which characteristics of movement patterns are determined by central or other factors is still a matter of debate (see chapter 1). In this thesis, studies have been presented which reveal properties of involuntary mechanisms involved in the coordination of stepping movements. This section will discuss the extent to which characteristics of the observed stepping patterns are determined by central or mechanical factors.

As discussed in chapter 2, the stereotypical spatio-temporal patterns observed during normal walking, seem to represent an optimal solution in regard to energy expenditure. However, given specific voluntary or mechanical constraints, this optimality is abandoned and other goals of coordination prevail. The studies of rhythmic walking and walking with visually guided step length (chapters 2 and 3) demonstrate that changes in voluntary control induce involuntary changes in central processes. As argued in chapter 2, these results indicate the existence of two independent central processes involved in the control of either frequency, or amplitude, of leg movements. In rhythmic walking, stepping patterns are produced which have distinct mechanical characteristics (i.e. contrary to normal walking, step length remains constant while step frequency increases). Consequently, the relative timing of the ensuing movement patterns differs from normal walking and walking with visually guided step length; during rhythmic walking, the relative duration of support and swing phases remains constant. Walking with visually guided step length also reflects involuntary changes in central control processes. The fact that in these conditions step length increases while step frequency remains constant reveals that the relative timing of phases within the stride cycle depends on amplitude of leg movements only.

In a somewhat different way than studies in preceding chapters, also chapter 6 illustrates that movement patterns with different mechanical characteristics can be brought about by voluntary control. This chapter demonstrates that voluntary adjustments of step length (i.e. making large or small steps in comparison to normal treadmill walking at different speeds) changes the mechanical conditions of walking. This leads to involuntary changes in kinematic patterns of pelvis and trunk which are predictable by mechanical models.

The studies of split-belt walking (chapters 4 and 5 and Ref. 1) show that when mechanical factors are manipulated (i.e. left and right leg are presented different belt speeds), leg movements immediately are adapted by involuntary mechanisms without the need for voluntary interventions. Learning effects (i.e. an adaptation of all gait parameters within 3 stride cycles, in stead of 12-15 stride cycles) can be observed when a particular split-belt walking condition is repeated[1]. These learning effects do not transfer to the contra-lateral leg in a similar, but inverted, split-belt condition. As argued in chapter 4 (and Refs. 1 and 2), the results of split-belt experiments suggest the existence of separate low level central coordinative mechanisms for both legs. These mechanisms are tuned by the proprioceptive feedback resulting from the mechanical characteristics of the movement patterns.

In chapter 5, the results of a dimensional analysis of spatio-temporal data of the stride cycle demonstrate that different mechanical conditions (i.e. different anthropometric characteristics in children and adults) and not differences in central processes (i.e. an immature nervous system in children) underlie differences in spatio-temporal characteristics of walking in adults and children older than four years.

It can be concluded that, in different ways, each of the presented studies illustrates that changing the mechanical conditions changes the spatio-temporal characteristics of walking. Different mechanical conditions not only result from different anthropometric characteristics or changes in external mechanical factors (e.g. split-belt walking), they can also directly or indirectly be caused by specific voluntary constraints. Thus, the studies show that the way involuntary coordinative mechanisms determine our movements depends on voluntary constraints and external mechanical constraints. When these constraints are minimal, human walking tends to be coordinated according to principles related to minimal energy expenditure. As illustrated by curvi-linear relationships between spatio-temporal parameters and a larger intra-subject variability in temporal parameters in children (chapter 5), the extent to which such principles actually govern observable kinematic patterns may depend on prerequisites for skilled movement. Such prerequisites include basic demands for movement production (for example, mobility and muscle force need to be sufficient), but also demands regarding coordination of movement; for instance, an effective interlimb and equilibrium control. It is likely that learning processes are important ingredients in both the development of basic demands for coordination, and skill acquisition.

When observing skilled movements, it is tempting to attribute specific features of these movements to central control mechanisms, or to assume different goals of central control processes in the production of these movements. However, it can be questioned whether typical features of the observed movement patterns in certain conditions must be attributed to central control mechanisms (as claimed by a motor programming perspective on human movement (see chapters 1 and 2), or whether they emerge from the nature of the complex interaction between the human body and the environment (as claimed by a dynamical systems approach). The results of this thesis suggest that when stereotypical characteristics of stepping patterns are observed, these must be attributed to mechanical factors, rather than central mechanisms. From the analysis of variant and invariant aspects of stepping patterns in the preceding section, it can be concluded that when stereotypes are observed these must not be confused with invariances. Therefore, a traditional motor programming view on movement behaviour (as described in chapter 1) cannot explain the results from the studies presented in chapters 2, 3, 4, and 5 (see also the discussion in chapter 2). Also the results presented in chapter 6 illustrate that mechanical aspects predominantly determine movement characteristics. This does, however, not imply that no programmed activity is involved in the observed movement patterns, but it illustrates the problems in defining what aspects of our movements are pre-programmed by central mechanisms.

Gait disturbances in Parkinson's disease

The two studies presented in chapter 3 demonstrate a fundamental deficit in regulating stride length in Parkinson's disease (PD) patients. The two studies show that, with exception of walking with visually guided step length, PD patients display linear relationships between amplitude, frequency, and speed. These relationships are similar to those of adults, but the amplitude of leg movements is consistently lower in PD patients. Walking with visually guided step length on the one hand shows that PD patients are able to substantially increase their step length. On the other hand, nearly all patients fail to do so with normal frequencies. Thus, the PD patients make large, but slow steps in these conditions. In regard to the involuntary mechanisms which control the timing of support and swing phases in the stride cycle in PD patients, the results presented in chapter 3 give evidence that these mechanisms are not different from healthy subjects. This lends further support to the claimed mechanical and not central determination of the relative timing of leg movements (see preceding section). It is very likely that the reduced stride length and the slower strides during walking with visually guided step length reflect the inability to perform fast movements in PD. Thus, this deviation from normal stepping patterns can be attributed to bradykinesia. However, it has been argued that the inability to initiate effective anticipatory postural activity may also be the cause for slower movements in PD[3,4]. Hence, the slowing down in PD can be interpreted as an effective adaptive behaviour of the central nervous system. This adaptive behaviour prevents balance problems due to changes in coordinative abilities (i.e. a defect in pre-programming). Although this might apply to patients with overt balance problems (for example, see chapter 3), it can be considered an unlikely explanation for the problems in regulating stride length in PD patients without overt balance problems. First of all, the fact that PD patients are able to voluntarily increase their stride length with (chapter 3 (study I) and Ref. 5) or without (chapter 3 (study II) and Ref. 6) external cues without encountering balance problems speaks against this. Secondly, studies of the effects of L-Dopa withdrawal versus walking with medication clearly demonstrate immediate effects upon particularly the amplitude of leg movements[7,8]. Contrary to this, postural control (i.e. postural reflexes[9,10] and anticipatory postural activity[11]) does not seem to improve by L-Dopa medication.

Despite these arguments for a fundamental deficit in regulating stride length in PD, other gait abnormalities of PD patients might possibly relate to problems in (anticipatory) postural control. In the patients who participated in the two studies presented in chapter 3, often typical Parkinsonian features of gait (e.g. problems initiating or continuing stepping; interference of cognitive activity with automaticity of stepping; hesitation when walking through a door opening or stepping over an obstacle) were observed. Despite the fact that these disturbances could be more incapacitating than the typically reduced stride length and walking speed, these problems all remained out-of-sight in the analysis of steady state gait during overground and treadmill walking. Chapter 3, and other studies[6,12], illustrate that PD patients are very well able to voluntarily control certain gait parameters. This ability can be a starting point for attempts to facilitate gait in PD[5,12-14], but it also may obscure some of the aforementioned gait problems in PD patients who are tested in relatively simple conditions, for instance, walking on a relatively short walkway or treadmill walking. Therefore, it can be questioned whether gait measurements in such conditions can reveal the most significant (i.e. the most disabling) gait disturbances in PD. Hence, additional methods are necessary to quantify and analyse these now covert gait disturbances. Such additional methods should be directed at the assessment of gait over longer periods (i.e. distances), and in functional conditions (for example, the combination of gait with cognitive tasks, or other motor tasks). Furthermore, studies of postural control during gait, or gait initiation and termination, may be more sensitive to specific gait disturbances in PD.

Clinical gait analysis

Normal versus abnormal movements

In clinical research or in routine based laboratory assessments which are made to support clinical decision making (e.g. biochemical or haematological measurements), the concept of "normal values" is implicitly or explicitly used when in certain test procedures the test values of a suspected patient are compared to the values of a population of normal subjects. This approach is also apparent in the criteria for clinical usefulness of movement analysis techniques which were stated in chapter 1. It can be questioned to what extent the concept of "normal movement" can be used in the clinical analysis of human movement. First of all, the concept implies that normal movements can be identified. However, most often "normal movements" are simply defined as the movements performed by healthy subjects. As illustrated in chapter 1, the coordinative mechanisms underlying these "normal movements" are only partly understood, and often different movement strategies can be used in order to achieve the same goal. Considering the fact that the characteristics of "normal movement" are under the influence of many often unidentified variables (e.g. anthropometric, psychological, and environmental variables), it is highly problematic to define normal movement when not all of these variables are accounted for in a comparison between healthy subjects and patients. Secondly, the "abnormal movements" of a patient need not directly be caused by a lesion at a particular level of the neuromusculoskeletal system, they can also be an effective adaptive behaviour of the central nervous system[3,4]. Thus, when observing the effects of pathology in the motor behaviour of a patient, a distinction must be made in signs and symptoms due to primary, secondary, or tertiary causes. In regard to neurological patients, primary disturbances are impairments directly caused by the neurological disease, e.g. spasticity or loss of selective control. Secondary musculoskeletal, or cardiorespiratory, problems (e.g. loss of strength, mobility, or endurance) may subsequently develop as a consequence of primary disturbances. Tertiary "abnormalities", however, are the different movement strategies caused by a voluntary or involuntary adaptation to the functional limitations which result from primary or secondary causes. Although this adaptation, for example, may lead to abnormal gait patterns, the adaptive behaviour may well be an effective strategy. In such cases, the comparison to so called "normal" gait patterns is meaningless.

To what extent can a contribution of human movement analysis to clinical decision making then be based upon comparisons between patients and normal values of a healthy population? Although a satisfying answer to these questions can not easily be given, the following demands seem minimal prerequisites for allowing such a comparison. First of all, important determinants of between subject differences in movement patterns of healthy subjects must be identified and understood. Secondly, influencing factors on normal movement patterns, or factors which determine the choice for certain movement strategies, must be identified. Thirdly, clinical research should be directed at defining specific impairments, their effects on motor functions, and effective movement strategies to avoid functional limitations due to specific impairments. The first two demands require that research is directed at the identification of determinants and influencing factors on human movement. These can both be internal (e.g. anthropometric or psychological characteristics) and environmental (e.g. different experimental conditions). When differential influences on movement parameters in different subjects are identified, these must be accounted for when between subject comparisons are made. This can, for example, be done by using appropriate normalisation methods, by matching individual characteristics, and by a standardisation of experimental methods and conditions. The third demand illustrates that a contribution of movement analyses to clinical decision making requires that clinical research efforts are directed at specific questions regarding the links between pathology, impairment, functional limitation, and disability (see also chapter 1). Not only is this important in the analysis of motor disturbances and the assessment of functional limitations and disability, but also in providing a rational basis for therapeutical interventions. Answering specific questions may require the time consuming development of specific methods of analysis, but it will sooner lead to a contribution of movement analysis techniques to clinical decision making than simply using available standard methods which are not specifically designed for analysing specific motor problems in patients.

It is hoped that, particularly in aforementioned respects, this thesis has made a contribution to clinical gait analysis. In the next sections, the value of the studies in this thesis respecting clinical gait analysis will be briefly discussed. These sections will particularly focus on implications of inter subject differences in amplitude, frequency, and walking speed, and different methodological aspects of clinical gait analysis. In a preceding section, it was argued that an adequate evaluation of gait disturbances in PD patients needs other methods of analysis in addition to those used in this thesis. Therefore, a last section will shortly indicate possible methods for the analysis and assessment of gait in functional conditions based on the methods used in chapter 6.

Amplitude, frequency, walking speed and other gait parameters

The studies presented in chapters 2, 3, and 6 illustrate that between subject differences in amplitude, frequency, and walking speed are important determinants of differences in other gait parameters. Although the importance of correcting for speed differences in comparing gait parameters between patients and healthy control subjects has been stressed[15,16], the results of this thesis suggest that amplitude and frequency rather than walking speed are major determinants of gait parameters. In this thesis, effects have been demonstrated on kinematic parameters only. However, amplitude or frequency of leg movements will certainly also affect other gait parameters. Obviously, changes in kinematic patterns coincide with changes in kinetic patterns, and different kinetic patterns can only follow from changes in the interaction between active and passive forces. Hence, also patterns of muscle activity can be expected to change when frequency or amplitude of leg movements is changed. The fact that the amplitude of leg movements determines the timing of support phases (see chapters 2 and 3), and the fact that, normally, patterns of muscle activity and angular displacement in lower leg joints are closely linked to onset and termination of support phases[17] suggest that at least the timing of these patterns will change with changes in the amplitude of leg movements. Obviously, amplitudes of angular displacement in hip, knee, and ankle will increase with increasing amplitude of leg movements. Depending on their function, specific effects of amplitude or frequency of leg movements on the amplitude of muscle activity may be expected. However, due to the complex nature of the relationship between electromyographical activity and mechanical effects of muscle activity[18], simple linear dependencies of muscle activity of either amplitude, or frequency, of leg movements should not be expected.

Thus, clinical gait studies in which deviating walking patterns are analysed not only should account for between subject differences in walking speed, but particularly for differences in stride length and frequency. Appropriate normalisation techniques should be used when these differences are caused by large anthropometric variations between subjects. Chapter 5 illustrates the usefulness of dimensional analysis in correcting for anthropometric differences. Contrary to different normalisation methods which have been used in other studies of human gait[19,20], this method provides a rational basis for correcting for between subject differences in anthropometric parameters. When differences in basic gait parameters are caused by other factors (e.g. functional limitations due to pathology), appropriate experimental protocols should be used. These protocols can be directed at studying gait parameters in dependence of either walking speed, stride length, or stride frequency, in order to establish whether these parameters are adapted similarly in different groups of subjects. Another possibility is to match these variables by means of a treadmill, or specific instructions. The next section will discuss to which extent a treadmill is a suitable instrument for this purpose.

Overground versus treadmill walking

Consistent differences in stepping patterns can be noted between overground and treadmill walking (see chapter 3 (study II), and Ref. 21). Although the cause of the typically reduced stride lengths during treadmill walking is not clear, a likely explanation can be found in the stationary surrounding during treadmill walking versus the optical flow during overground walking. Studies of the effects of optical flow manipulations have shown that these affect the relationship between stride length and frequency[22-24]. However, other factors may also play a role. For example, particularly when the dimensions of the treadmill walking surface are small, subjects may be cautious and avoid making large steps. Furthermore, contrary to overground walking, voluntary control during treadmill walking is directed at maintaining a certain position on the treadmill rather than controlling walking speed. This difference with overground walking may influence the relationships between walking speed, and amplitude and frequency of leg movements. Therefore, it can be questioned whether walking on a treadmill is suitable for studying human walking. The answer to this question obviously depends on the purposes of a study. When mechanical aspects of walking are studied, a treadmill which does not display speed fluctuations should yield similar results to overground walking. Other biomechanical differences, i.e. the absence of wind drag, can be neglected[25]. However, typical changes in amplitude and frequency of stepping movements should not be ignored when the results of treadmill walking studies are generalised to walking in general. Furthermore, care should be taken that subjects are familiarised with treadmill walking before measurements are made[26]. It has been reported that, even after at least 15 minutes of treadmill walking, patterns of muscle activity are slightly higher than in overground walking[27]. These results were obtained for identical walking speeds, and stride lengths and frequencies. Thus, independently from the effects of the typically reduced stride length during treadmill walking, neurophysiological parameters of gait may be slightly different in comparison with overground walking at a similar speed.

An appropriate treadmill (i.e. a treadmill with minimal speed fluctuations and a large walking surface) is, of course, a very convenient instrument for studying aspects of gait. Yet, validation studies are necessary before the results of treadmill walking studies can be generalised to overground walking. This may be particularly true when gait disturbances are investigated by means of a treadmill. In regard to clinical gait analysis, a treadmill can only be used for analysing aspects of gait disturbances. When the functional quality of gait in a patient is to be assessed, a treadmill obviously is not a suitable instrument.

Mean walking patterns versus a cycle-to-cycle approach

In this thesis, the established relationships between spatio-temporal parameters of walking always represent the relationships between individual, or group, *mean* values. Whenever possible, the intra-individual variability in spatio-temporal parameters (i.e. the variability from stride cycle to stride cycle) was considered, but, due to limited walking distances during overground walking (see chapters 2 and 3), a systematic analysis of this variability was only possible in treadmill walking conditions. Chapter 5 demonstrates that such measures of variability may be of value in the assessment of gait. The models presented in chapter 6 elucidate the importance of successive foot placements in controlling the centre of mass trajectory during walking. Thus, these models suggests that a cycle-to-cycle control of the centre of mass trajectory will be reflected in variability in spatio-temporal parameters of the stride cycle. A recent study[28] shows that stride durations during longer periods of walking have fractal characteristics. These normally unobserved fractal characteristics of human locomotion may be related to (higher) neural mechanisms involved in the control of walking rhythm. In addition to aforementioned studies, other studies in which gait parameters are studied from cycle-to-cycle reveal more about the nature of movement control in specific conditions. Examples of such studies are the cycle-to-cycle approach applied in a study of the immediate adaptation to split-belt walking conditions[1], studies of gait initiation and termination[29], and studies of strategies used in obstacle avoiding[30,31]. All aforementioned studies suggest that properties of coordinative mechanisms, and perhaps of central control processes, can be identified by a cycle-to-cycle approach. Thus, despite the fact that an analysis of mean gait patterns can be very useful in the assessment of gait disturbances, studying the adaptation of individual movement patterns may yield more information about specific disturbances. Therefore, such an approach may be important in clinical gait analysis.

Assessment of gait in functional conditions

As argued in preceding sections (and in chapter 1), the assessment and analysis of gait disturbances should not only be directed at a stereotypical "mean gait pattern" obtained in a standardised laboratory environment. The studies in this thesis illustrate that, in order to understand gait disturbances, analyses should also be directed at the adaptability of gait patterns in different conditions, and inter and intra subject variability of gait parameters should also be considered. In regard to gait disturbances in PD, a preceding section argued that additional methods are necessary which allow the study of gait in varying functional conditions. The analysis of overground walking over longer periods (distances) than allowed by typical walkway measurements of gait requires the availability of measurement systems which allow the recording of kinematic and physiologic signals in real life situations. Recent developments have lead to new possibilities of ambulatory measurements of movement by means of accelerometers[32]. When multiple accelerometers are used, a complete analysis of body movements now seems possible[33,34]. However, this approach requires the use of an extensive instrumentation and several methodological problems have to be overcome. Thus, in addition to this approach, relatively simple methods for gait analysis which require minimal accelerometers are desirable. Chapter 6 suggests that trunk acceleration patterns can be a meaningful target of analysis. A recent study[35] indeed indicates that, with relatively simple methods, measures of variability in trunk acceleration patterns can be obtained which are indicative of the stability of walking. Particularly in gait disturbances which result from balance disorders, an increased variability of trunk movements might be indicative for the risk of falling.

The methodology used in chapter 6 provides a basis for gait assessment in different functional conditions. Applied to accelerometry data obtained by ambulatory measurements, it could be useful in investigating the effects of specific conditions, or task variants, on gait in healthy subjects and patients. As the presented models demonstrate the biomechanical mechanisms which underlie changes in patterns of linear pelvic displacement, the influence of changes in walking speed, stride length, and stride frequency on amplitude and timing of linear displacement of the pelvis can be understood. As argued in preceding sections, this is important for clinical gait analyses in which normal and abnormal walking patterns must be differentiated. By means of a harmonic analysis of trunk kinematics, aspects of deviating walking patterns (i.e. abnormal amplitude and timing of trunk movements, increased asymmetries, and an increased variability of movement patterns) can be quantified. Based on the models presented in chapter 6, an analysis of accelerometry data of the pelvis can perhaps also be used to estimate stride length and frequency in situations outside a laboratory environment.

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Summary

This thesis explores characteristics of human walking in dependence of walking speed, stride length, or stride frequency. A number of studies were performed in different groups of subjects (i.e. children between four and ten years old, young adult subjects, healthy older subjects, and Parkinson's disease (PD) patients), and in different conditions (i.e. overground and treadmill walking at different speeds, voluntary adaptation of walking to external cues, split-belt walking, and voluntary adaptation of stride length (and frequency) during treadmill walking at different speeds). The studies were performed in order to document variant and invariant aspects of stepping patterns, to identify characteristic between group differences, and to investigate which factors explain variant and invariant aspects of stepping patterns and which factors explain differences between groups.

The results of the different studies demonstrate that in all subjects stereotypical relationships between spatio-temporal parameters of the stride cycle can be observed in each of the different conditions. However, the nature of these relationships changes in different conditions, and invariant relationships between spatio-temporal parameters of the stride cycle were not observed. The only exception to this is the maintenance of an alternating pattern of stepping movements. It is shown that, rather than walking speed, amplitude and frequency of stepping movements are major determinants of other characteristics of human walking. Thus, stride length and frequency can both be a source of between subject and between group differences in gait parameters. A number of specific effects of either amplitude or frequency on other gait parameters have been identified.

Despite the existence of characteristic between group differences, condition specific effects on the relationships between walking speed, stride length, and stride frequency are highly similar between the different groups of subjects. Except for the typically reduced stride length and walking speed, stepping patterns in PD patients without major postural problems are similar to those of age-matched subjects. Observed problems in stride length regulation probably reflect an inability to perform fast movements (i.e. bradykinesia) in PD. Although large differences in spatio-temporal data were observed between children and adults, these differences no longer exist when differences in anthropometric variables are taken into account by means of a dimensional analysis. However, intra and inter subject variability of gait parameters is consistently higher in children.

The different studies show that differences in spatio-temporal gait parameters between subjects, groups, and conditions, often can be explained by differences in mechanical factors. Changes in the mechanical characteristics of walking patterns can be brought about by different anthropometric characteristics, changes in external mechanical conditions, or specific voluntary constraints. It is concluded from the presented studies that mechanical rather than central factors determine the way in which stepping movements are adapted during different walking conditions. In regard to the control of frequency and amplitude of stepping movements, the studies suggest the existence of different central control mechanisms. The studies illustrate that knowledge of the way gait parameters are determined by amplitude and frequency of stepping movements, or walking speed; knowledge of the relationships between these three basic gait parameters; and knowledge of factors influencing their relationships, are essential in understanding normal and deviating gait patterns.

Samenvatting

In dit proefschrift worden karakteristieken van het looppatroon van de mens bestudeerd in afhankelijkheid van loopsnelheid, stapgrootte en stapfrequentie. Het algemene doel van dit proefschrift is het bijdragen aan de analyse en interpretatie van loopstoornissen als gevolg van pathologische veranderingen in het centrale zenuwstelsel. In een inleidend hoofdstuk wordt betoogd dat hiertoe meer begrip van de rol van het centrale zenuwstelsel in het bepalen van zowel normale als afwijkende karakteristieken van stappatronen noodzakelijk is. Nadat in dit eerste hoofdstuk de algemene achtergronden van het voor het proefschrift verrichte onderzoek zijn geschetst worden in vijf volgende hoofdstukken de resultaten van verschillende studies gepresenteerd. In deze studies worden de looppatronen van kinderen, jonge volwassenen, ouderen en Parkinsonpatiënten bestudeerd in verschillende omstandigheden. De studies hebben als specifiek doel het vastleggen en, waar mogelijk, verklaren van variante en invariante aspecten van stappatronen, en het vaststellen en verklaren van karakteristieke verschillen in stappatronen tussen groepen van proefpersonen.

In hoofdstuk 2 worden veranderingen in de looppatronen van gezonde volwassen proefpersonen die de opdracht krijgen hun looppatroon willekeurig aan te passen aan externe stimuli vergeleken met de veranderingen in het looppatroon tijdens lopen op verschillende snelheden. De willekeurige aanpassing van het looppatroon aan de externe stimuli (d.w.z. de ritmen van een metronoom, of dwars op het loopoppervlak aangebrachte strepen) leidt tot het vastleggen van stapgrootte of stapfrequentie. De resultaten tonen dat de relaties tussen loopsnelheid, stapfrequentie en stapgrootte veranderen. Bij lopen op verschillende snelheden bestaan individueel karakteristieke relaties tussen deze drie parameters; een toename in loopsnelheid wordt steeds bereikt door een toename in zowel stapfrequentie als stapgrootte. Tijdens lopen op ritme blijft stapgrootte min of meer constant als in opeenvolgende metingen de stapfrequentie toeneemt. Tijdens het stappen over strepen blijft daarentegen stapfrequentie constant als stapgrootte toeneemt. In elk van de drie condities blijkt de timing van stand- en zwaaifasen te worden bepaald door stapgrootte. De onwillekeurige veranderingen van het looppatroon in de verschillende condities kunnen niet worden begrepen uit de aard van de opdrachten in de desbetreffende condities. De resultaten suggereren het bestaan van verschillende centrale sturingsmechanismen voor de amplitude en frequentie van stapbewegingen, de timing van stand- en zwaaifasen lijkt daarentegen niet centraal bepaald.

In hoofdstuk 3 worden twee studies gepresenteerd waarin de loopstoornissen van Parkinsonpatiënten worden bestudeerd in vergelijking tot de looppatronen van gezonde proefpersonen van eenzelfde leeftijd. In een eerste studie worden de veranderingen in de looppatronen van Parkinsonpatiënten bestudeerd als deze de opdracht krijgen hun looppatroon willekeurig aan te passen aan externe stimuli. In een tweede studie worden veranderingen in het looppatroon tijdens vrij lopen op verschillende snelheden vergeleken met de veranderingen tijdens lopende band lopen op verschillende snelheden. Uit de resultaten blijkt dat de meeste patiënten in staat zijn hun looppatroon aan te passen in overeenstemming met de verschillende opdrachten. Met uitzondering van kenmerkende verschillen in stapgrootte zijn stappatronen van Parkinsonpatiënten zonder balansproblemen in de onderzochte condities vergelijkbaar aan die van gezonde leeftijdgenoten. De timing van stand- en zwaaifasen is bij Parkinsonpatiënten niet afwijkend wanneer rekening wordt gehouden met een afgenomen stapgrootte. De bij de patiënten waargenomen veranderingen in het reguleren van de stapgrootte lijken te kunnen worden toegeschreven aan een onvermogen tot het uitvoeren van snelle bewegingen.

In hoofdstuk 4 worden onwillekeurige aanpassingen van het looppatroon bestudeerd bij gezonde jonge volwassenen die aan het linker- en rechterbeen verschillende snelheden

opgelegd krijgen door een lopende band. Dit zg. split-belt lopen wordt vergeleken met het lopende band lopen op verschillende snelheden. Uit de resultaten blijkt dat het split-belt lopen wordt gekenmerkt door veranderingen in de duur van de loopcyclus, een asymmetrie in stapgrootte, en asymmetrische veranderingen in de timing van stand- en zwaaifasen. Alleen de timing van de dubbele standfasen wordt links en rechts gelijk aangepast. Ondanks grote verschillen in linker- en rechterbandsnelheid zijn alle proefpersonen in staat hun looppatroon binnen een beperkt aantal loopcycli automatisch aan te passen. De veranderingen in het looppatroon tijdens split-belt lopen wijzen op een variante timing van stand- en zwaaifasen.

In hoofdstuk 5 worden de resultaten gepresenteerd van een studie naar veranderingen in de looppatronen van gezonde kinderen tussen vier en tien jaar oud tijdens lopende band lopen op verschillende snelheden en tijdens split-belt lopen. Door middel van een dimensieloze analyse die rekening houdt met verschillen in lichaamslengte worden de stappatronen van de kinderen vergeleken met die van jonge volwassenen. De analyse toont aan dat verschillen in stappatronen tussen volwassenen en kinderen kunnen worden verklaard door verschillen in lichaamslengte. Stapparameters van kinderen van vier jaar en ouder blijken identiek te worden aangepast als bij volwassenen. Bij kinderen is wel sprake van een hogere intra- en inter-individuele variabiliteit van stapparameters dan bij volwassenen.

In hoofdstuk 6 wordt het verplaatsingspatroon van het bekken tijdens lopende band lopen bestudeerd bij een groep jonge volwassenen. Naast het spontane looppatroon tijdens lopen op verschillende snelheden wordt op elke bandsnelheid ook het lopen met grote en kleine passen bestudeerd. Door middel van eenvoudige mechanische modellen van de lichaamszwaartepunt-beweging tijdens lopen worden voorspellingen gemaakt van de grootte en timing van de bekkenbeweging. Uit de resultaten blijkt een hoge mate van overeenstemming tussen model-voorspellingen en werkelijke verplaatsing van het bekken. De grootte van de links-rechts bewegingen van het bekken is afhankelijk van schredefrequentie en stapbreedte. De grootte van de verticale beweging van het bekken wordt bepaald door stapgrootte. De voor-achterwaartse beweging van het bekken wordt bepaald door een combinatie van stapgrootte en stapfrequentie.

Het afsluitende hoofdstuk 7 bevat een algemene discussie van het in dit proefschrift gepresenteerde werk. De verschillende studies tonen aan dat bij alle proefpersonen stereotype veranderingen van stapparameters kunnen worden waargenomen in de verschillende condities. De relaties tussen stapparameters varieren in verschillende omstandigheden. Met uitzondering van de handhaving van een alternerend patroon van beenbewegingen kunnen geen invariante relaties tussen tijd-afstand parameters van de loopcyclus worden waargenomen. Uit de resultaten blijkt dat stapgrootte en stapfrequentie belangrijke determinanten van andere loopparameters zijn. Ondanks het bestaan van karakteristieke groepsverschillen zijn specifieke effecten van de verschillende condities op de relaties tussen loopsnelheid, stapfrequentie en stapgrootte zeer vergelijkbaar tussen de groepen. De in dit proefschrift gepresenteerde studies tonen aan dat verschillen in tijd-afstand parameters van het looppatroon tussen proefpersonen, groepen, of condities vaak kunnen worden verklaard door verschillen in mechanische factoren. Veranderingen in de mechanische karakteristieken van looppatronen kunnen tot stand komen door verschillende anthropometrische karakteristieken, veranderingen in externe mechanische condities, of specifieke willekeurige veranderingen van het looppatroon. Uit de studies kan worden geconcludeerd dat vooral mechanische factoren en niet centrale sturingsmechanismen de aard van de aanpassing van stapparameters in verschillende loopcondities bepalen. De studies illustreren dat kennis van de manier waarop loopparameters worden bepaald door stapgrootte, stapfrequentie of loopsnelheid, kennis van de manier waarop deze drie basale stapparameters met elkaar samenhangen, en kennis van factoren die deze samenhang beïnvloeden, essentieel zijn in het begrijpen van normale en afwijkende looppatronen.

Nawoord

Dit proefschrift werd mogelijk door een belangstelling voor beweging, bewegingssturing en bewegingsstoornissen die wordt gedeeld door Prof. Rispens, hoofd van de Werkgroep Bewegingswetenschappen, en Prof. Lakke, tot enkele jaren geleden bijzonder hoogleraar in de extrapyramidale ziekten aan de RijksUniversiteit Groningen. Beiden hebben deze belangstelling ondermeer tot uitdrukking gebracht in hun bijdrage aan het tot stand komen van een Laboratorium voor Houdings- en Bewegingsanalyse in het Academisch Ziekenhuis te Groningen. Dit in de wandelgangen "Looplab" genoemde laboratorium dient als gedeelde werkplek voor onderzoek naar motorisch functioneren door de klinische afdelingen Revalidatie, Neurologie en Orthopaedie.

Hoewel het in dit proefschrift gepresenteerde werk slechts ten dele in het Looplab plaatsvond zijn de verrichte studies wel alle uitgevoerd vanuit een bij het Looplab passend perspectief op onderzoek aan menselijk bewegen. Een aan het begin van dit onderzoeksproject gesteld doel was het door middel van een draagbaar meetsysteem analyseren van loop- en balansstoornissen bij Parkinsonpatiënten. Dat dit doel niet in dit proefschrift is vormgegeven heeft twee redenen; een fors vertraagde levering van het te gebruiken meetsysteem en de noodzaak van het oplossen van enkele fundamentele problemen bij het gebruiken van versnellingsopnemers voor het analyseren van beweging. Gelukkig zijn inmiddels vorderingen geboekt bij het ontwikkelen van een methode voor het analyseren van het werkelijk vrije gaan. Mede daarom hoop ik dat dit proefschrift een tussenstand zal blijken in mijn onderzoek aan de loop- en balansfunctie van ouderen en Parkinsonpatiënten. De belangrijkste redenen voor mijn hoop dat dit proefschrift een tussenstand zal blijken zijn echter het plezier dat ik heb gehad aan het nu afgeronde onderzoek en mijn wens om door verder onderzoek bij te dragen aan het oplossen van klinische vragen en problemen met betrekking tot het menselijk bewegen.

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About the author

After graduating at the 'Academie voor Fysiotherapie' in Leeuwarden in 1985, Wiebren Zijlstra worked as a physiotherapist in Eksjö and Nassjö in Sweden. From 1986 to 1991, he attended the study Human Movement Sciences at the 'RijksUniversiteit Groningen' and combined this with working as a physiotherapist in Groningen and Sweden. After graduation, a fellowship from 'het Prinses Beatrix Fonds' offered the opportunity to work in the laboratory of Prof. Dr. V. Dietz and Prof. W. Berger at the 'Neurologische Klinik der Albert-Ludwigs-Universität' in Freiburg, Germany. In July 1992, he started to work as a research assistant, and from 1993 to 1997 as a PhD student, for the 'Werkgroep Bewegingswetenschappen' / the Department of Neurology of the 'RijksUniversiteit Groningen'. During this time, he conducted studies on human locomotion in the laboratory of human movement analysis of the University Hospital Groningen and the 'Motoriklabor' in Freiburg. At present, he is employed as a researcher at the 'Werkgroep Bewegingswetenschappen' / the Department of Rehabilitation.