DIFFERENCES BETWEEN MACHINE AND HUMAN TESTING OF SHOCK ABSORBING SYSTEMS

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To Ana and Toni for their endless patience and ever present support

A Ana, incondicional companya en aquest viatge i en quants pugi somniar

A mon pare i ma mare que hem regalaren aquest mon i en especial a mon pare que ja no està
A Toni l'incansable que manté viu al xiquet que s'amaga dintre de mi
... i en especial a la gent que lluita per allò en que creu.

Abstract

This thesis documents a study on the sources of the differences found between results from machine and walking testing of shock absorbing systems. A complex programme of experiments was conducted at the Institute of Biomechanics of Valencia to explore the four most outstanding statements proposed with this respect:

1.- No accurate simulation of impacts by machine test. This was investigated by comparing results from testing materials simulating impact forces with results from walking tests.

2.- In use materials degrade and their properties change and existing machine testing methods could not replicate material properties during walking. A new testing method was developed to measure the recovery ability of materials by simulating plantar pressures and results compared with walking tests.

3.- Shoe effect on walking kinematics and heel pad confinement has greater influence on shock absorption than material properties. An instrumented pendulum was developed to study the heel pad. Insole materials were evaluated in walking tests, in pendulum tests and in different machine testing including the new method developed simulating plantar pressures and the results compared.

4.- Accommodation to impact conditions occurs according to a controlled proprioceptive feedback model. Accommodation, impact perception, comfort, walking and passive biomechanical variables and material properties were studied in relation to system's input, output and goal.

Accurate simulation of impacts improved the ability of machine test to predict the walking performance of materials, but not upper body shock transmission. Properties of materials such as recovery ability, stiffness and hardness play an important role in concepts and passive interaction but mainly by influencing accommodation. Accommodation was identified as the source of differences of results between machine and walking tests of shock absorbing materials. The human body was described as comprising two independent mechanical systems: One system, governed by the elasticity and hardness of materials, it is defined by impact forces and accelerations that are inversely related to upper body transmission and control the perceived impact through foot position and knee bend. The other system is defined by heel pad stiffness, insole properties at initial loading and passive interaction that regulate upper body shock transmission by ankle inversion for comfort control. Passive interaction is defined in this thesis as the mechanical coupling between insole and heel pad that determines the properties of the system either through heel pad confinement or compression. Machine tests appear to predict results with respect to the first system but not the second, which required passive human testing.

For insole use, high-energy absorption materials are preferred. These are capable of increasing elastic deformation to reduce impact forces and accelerations without increasing initial-maximal stiffness by passive interaction thus avoiding any increase of head transmission due to accommodation. Heel pad properties were described by three mechanical components accounting for 93.08% of total variance: These are an elastic component, a viscoelastic component and a component related to elastic deformation at low stiffness. Differences were found between shod and barefoot test results. With barefoot there was an initial low stiffness (18-50 kNm^{-1}) response that was not evident in the shod tests which showed elastic deformation related to final stiffness. With barefoot, the elastic component accounted for impact forces variance (> 70%) and initial deformation component for peak force time (> 60%), while shod impact forces were related mainly to the elastic deformation component (> 60%) whereas rate of loading and acceleration were related to the initial-maximal stiffness component (>20%).

Differences in heel pad mechanics due to age, gender and obesity were observed. Although the heel pad properties degraded with age, losses appeared to be compensated by obesity.

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1. INTRODUCTION

1. Introduction

This thesis focuses on the biomechanics of shock absorption during the human walking process. Shock absorption refers to absorption and attenuation of impact forces that occur when the feet contact the ground during walking, running or jumping. These impact forces transmit a shock through the body from heel to head via the skeletal structure. Impact forces of three times body weight and accelerations to 17 g (gravity units) have been registered during running.

Degenerative joint diseases, low back pain and stress fracture are some of the health problems commonly attributed to the impact forces and shock waves that occur during walking. It has been demonstrated that these health effects, and ease and comfort of movement depend significantly on the magnitude of the forces and the frequency of the shock waves. Contradictory, walking, running and jumping are important stimuli for growth and musculoskelatal development and maintenance of fitness in later life.

Judging by the amount of relevant scientific literature published over the last 20 years, the study of the biomechanics of shock absorption has been extensive with significant developments in clinical practice. These include the treatment of sports injuries, fracture healing and bone repair, orthopaedics, elderly care, child growth patterns, rehabilitation and of footwear design.

The overall direction of that research was to establish how best to protect the "fragile" body (Robbins & Gouw, 1990) from the effects of impact loads and subsequent shock waves yet, almost twenty years on, the incidence of impact related problems show n o sign of decreasing. More recent research has shown that if the effectiveness of shock absorbing footwear materials is tested both by machine methods and by methods using volunteers wearing these footwear materials, the results are different and this has led to several authors questioning the accepted concepts of shock absorption. Many explanations have been proposed for these differences but there is no consensus of current scientific opinion except to agree that the role of individuals is increasingly relevant to the understanding of shock absorption.

The human body has been described as a complex feedback control system for shock absorption (Forner et al. 1995; Robbins & Gouw, 1991). The body possesses highly effective natural systems to reduce and control the effects of impact loading and described, by some authors, as significantly more effective than the artificial systems. The alternative view is that the interaction between footwear and the body's natural system serves to enhance the body's ability to absorb shock loads. This interaction is called accommodation.

Understanding the differences between the natural and artificial systems and especially the mechanism of accommodation will significantly advance the study of shock absorption and accelerate improvements in footwear design, promote the development of new materials, rehabilitation methods and orthopaedic practices and advance these aspects of biomechanics. This, in turn, would encourage the development of new fields of study such as impact perception, natural shock absorbers and accommodation and put new energy in shock absorption analysis.

The main reasons that have been advanced to explain the discrepancies between human and machine testing are:

i.- Current machine testing does not accurately simulate impact forces,

ii.- Materials degrade with use showing inferior properties to those measured by machine testing,

iii.- The effect of the shoe in walking kinematics and that heel pad confinement has greater influence on shock absorption than the materials used,

iv.- Accommodation to impact conditions.

The research investigates these statements; each related to a different research field in biomechanics and the thesis describes the many issues that were addressed and evaluated. A number of questions were formulated in relation with each statement, the hypotheses to be tested were established and the null hypothesis considered. A programme of experiments devised to address each statement extended from the straightforward to the complex was devised. The experimental programme is outlined below.

➊ *Current machine testing does not accurately simulate impact forces.* This statement refers to research on machine testing of materials from a biomechanical point of view. It was considered that testing materials simulating the magnitude and duration of walking impact forces would yield a good prediction of material's performance during walking as given by ground reaction forces and accelerations. To test this hypothesis a new machine testing method previously developed at the IBV was used to simulate impact loading and more accurately measure properties of a sample of materials. The results were compared with forces and accelerations registered in walking tests with the same materials[1]. If no acceptable agreeement was observed between machine

[1] This work was published in Foot and Ankle in 1995 (see Forner et al., 1995)

testing and human testing the following statements were considered:

- That testing materials by simulating impact forces do not predict their performance during walking
- That the source of differences between human testing and machine testing of shock absorbing systems should be related to the other research issues considered in this thesis.

❷ *Materials degrade with use showing inferior properties than measured by machine testing.* This statement relates to long term properties of materials in use. It has been suggested that the recover-ability of materials due to residual displacement after removing an applied compressive load, when measured using a machine test simulating walking underfoot pressures, was related to long term performance of materials. To explore this statement it was assumed that materials showing higher residual displacement (worse recovery) would perform worse during walking giving rise to greater and higher frequency impact forces and higher accelerations. To test both hypotheses, a new machine testing method was developed simulating plantar pressure loading to measure the recover-ability of a sample of materials[2]. The residual displacement was then compared with results from long term subjective walking tests and shock absorption testing during walking. If no relationship was observed the following statements were considered:

- That degradation of materials in use cannot be predicted by measuring the residual displacement and would require further research,
- That residual displacement as a measure of a material's degradation in use does not account for differences between human and machine testing of shock absorbing systems,
- That these differences should be related to the remaining research issues considered in this thesis.

❸ *The effect of the shoe in walking kinematics and heel pad confinement has greater influence on shock absorption than the materials used.* This statement is related both to human gait analysis and to heel pad biomechanics. It was assumed that if differences were found in impact forces and accelerations registered by humans walking with different insoles in the same shoe, these differences were attributable to the insole materials. If this was the case, it was considered that materials' influence in

[2] Two papers on this work have been accepted for presentation in the 2001 Footwaar Symposium, to be held in July in Zürich

shock absorption could be due to their role as shock absorbers, their influence in human accommodation or to passive interaction. Passive interaction is a new concept related to the inter-action between man and insole material (described in this thesis) and refers to the underfoot properties resulting from mechanical coupling between heel pad and insole.

Concerning these possibilities, passive impact tests were conducted using a ballistic pendulum developed for this research. Shod humans wearing different insoles participated in this experiment. Any differences that persisted between insoles were attributed to either materials properties or passive interaction. Otherwise they were considered to be related to accommodation.

The human heel pad is an inherent shock absorber and it is important to understand the mechanics of energy absorption and the inter-action of heel pad and insole materials. To do this it was necessary to develop a method for testing heel pad properties. To assess the effect of gender, age and obesity on heel pad mechanics, initially pendulum tests were conducted on barefoot participants[3].

Finally, correlation analysis between machine tests of materials and both passive and active human tests was conducted to assess the relative contribution of materials' properties and passive interaction.

❹ *Accommodation to impact conditions.* This statement is related to human perception of impact forces and accommodation and described in Chapter 4.

Impact forces and acceleration from walking tests were analysed to assess whether accommodation had occurred. If any reverse relationship was found between impact forces and accelerations then accommodation would be considered to have occurred. At the same time, results from pendulum test of shod humans and from walking tests were compared and any differences would be attributed to accommodation. Many questions were explored in relation to this statement.

The human body has been described as a feedback model [Chapter 4] and several experiments were completed to evaluate its different components. These components were input variables, the perception of impacts, the objectives of the system and system output (feedback) that modifies undesirable stimuli [Chapter 5]. A sample of ten insoles was chosen from a greater set according to results of different machine testing methods. Ten participants tested these insoles in shoes and various biomechanical and subjective variables measured that were then analysed according

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[3] The development of the pendulum and the results of the analysis of sex, age and obesity influence in heel pad properties are included in e paper sent for publication in 2000 to the Journal of Applied Biomechanics

to the experimental programme.

1.1. Thesis structure

The thesis is structured in eleven Chapters. In the first four the state of current knowledge is presented and related terminology defined. Chapter 2 is dedicated to shock absorption by the human body during walking, and explains the underpinning knowledge and defines the terminology. It also describes impact forces, their effect on the human body and different means of absorbing shock and a review of heel pad anatomy and biomechanical properties. Further, it includes a thorough review of literature associated with analyses of heel pad shock absorbing properties. Chapter 2 compares results between machine testing and human testing of shock absorbing systems and presents a case to support the main objective of this thesis to explain the discrepancies that are in detail described in Chapter 3. Chapter 4 is devoted to the theory of the proprioceptive model and accommodation along with a literature survey of current investigations in impact perception, as these are the newer and more complicated concepts, which raise many questions on the subject of shock absorption.
The rationale of the experimental programme and details of experimental design are discussed in Chapter 5. The experimental work is presented in Chapters 6 to10 and is self-contained insofar as the material and methods used for each experiment are described together with results, discussion and conclusions.

Chapter 6 compares results from a machine testing method that simulated impact loading during walking with walking tests to explore the first research issue. Chapter 7 describes the instrumented pendulum test developed for heel pad testing and assesses the influence of gender, age and obesity on heel pad biomechanics. Chapter 8 describes the selection processes for insole materials to be used in the next experimental phase that necessitated the development of a new testing method to simulate plantar pressure loading. The walking tests described in Chapter 9 are divided into objective and subjective techniques, the former include shock absorption, pressure distribution and rear-foot movement whilst subjective techniques include comfort and impact perception. Chapter 10 describes experimental pendulum testing on shod people to assess passive impact properties.

Chapter 11 is structured in four parts, one for each of the statements. The issues implicit in the research objectives are explored and some answers proposed, results discussed and conclusions presented. The final conclusions of this research -including recommendations for further research - are also included at the end of the Chapter.

The structure of the thesis is shown schematically in Figure 1.1

1.2. Aims and Objectives

The aim of this work was to analyse the reasons for the discrepancies between the results of machine testing and human testing of shock absorbing materials. The objectives were to investigate and test the biomechanical and methodological aspects of each of the following statements. Considering each statement in turn:

1. "Current machine testing methods do not accurately simulate impact forces".
It was considered that machine testing insole materials simulating the magnitude and duration of walking impact forces would yield a good prediction of materials' performance during walking as given by ground reaction forces and accelerations.
The work carried out to test this hypothesis included the use of *a new machine testing method developed at IBV* to accurately simulate impact loading. The results of this test are compared with walking tests to check if accurately simulation of impact loading yields results in agreement with walking results.

2. "Materials degrade in use showing lower properties than measured by machine testing".
Two hypotheses were considered in relation to this statement:

i. That the recovery-ability of materials due to the residual compressive displacement measured using a machine test simulating walking underfoot pressures is related to long term performance of materials.
ii. That those materials showing higher residual displacement during walking give rise to greater and faster impact forces as well as higher accelerations.

These hypotheses were tested by comparing data from walking tests with data from a new testing method simulating plantar pressure loading and measured the recovery-ability of materials.
For this it was necessary:

- To develop a new machine testing method simulating plantar pressure loading to measure the recover-ability of a sample of materials.
- To identify properties of material insoles that describe recovery-ability and its role in shock absorption.
- To compare results from the new method with results from long term subjective and objective walking tests.

3. "The effect of the shoe in walking kinematics and heel pad confinement has greater influence on shock absorption than insole materials".

It has been proposed that if differences were found in impact forces and accelerations registered from human walking with different insoles inserted in the same shoe these differences were due to materials' influence. This could be due to either their role as shock absorbers or their influence in human accommodation or passive interaction.

This was studied by testing ten insoles in the same shoe type in active (walking) and passive (pendulum) tests.

If differences between insole materials were found in active and in passive tests they were attributed to either materials properties or passive interaction. If they appeared only during walking they were considered related to accommodation.

To assess the role of insole materials, their performance in passive and active tests were compared using shod and barefoot human volunteers.

For this it was necessary to:

- Develop a method for testing and studying heel pad mechanics of barefoot and shod humans
- Analyse the influence of gender, age and obesity in heel pad mechanics
- Analyse the effect of insole material in passive impact attenuation
- Analyse the effect of insole material in walking kinematics and heel pad confinement and its role in shock absorption.
- Correlation analysis between machine tests of materials and both passive and active human tests.

4. **"People accommodate to impact conditions".**

The human body has been described as a feedback model [Chapter 4]. This statement required complex analysis of the different aspects of impact perception, input and output, as well as system's goal. Initially it was necessary to assess whether accommodation took place and, if so, then conduct a series of experiments to establish the different components of the model analysing the results from objective and subjective walking tests, pendulum and machine tests on ten different insoles.

For this it was necessary to:

- Develop a new machine testing method simulating plantar pressure loading.
- Analyse accommodation to impact conditions.
- Test the theory of plantar discomfort at accommodation onset.
- Analyse the response of the proprioceptive model to kinematics adjustments.

- Study the role of biomechanical, human and insole material variables in proprioception input.
- Validate the proprioceptive model for studying the role of comfort and biomechanical variables.
- Analyse impact perception with respect to the following:
    - Role of impact perception in shock absorption
    - Sensitivity of humans to perceive impacts
    - Comparison of impact perception in passive and active tests.
    - To assess the value of a reference condition for impact perception rating
    - Comparison of perception in two different active tests.

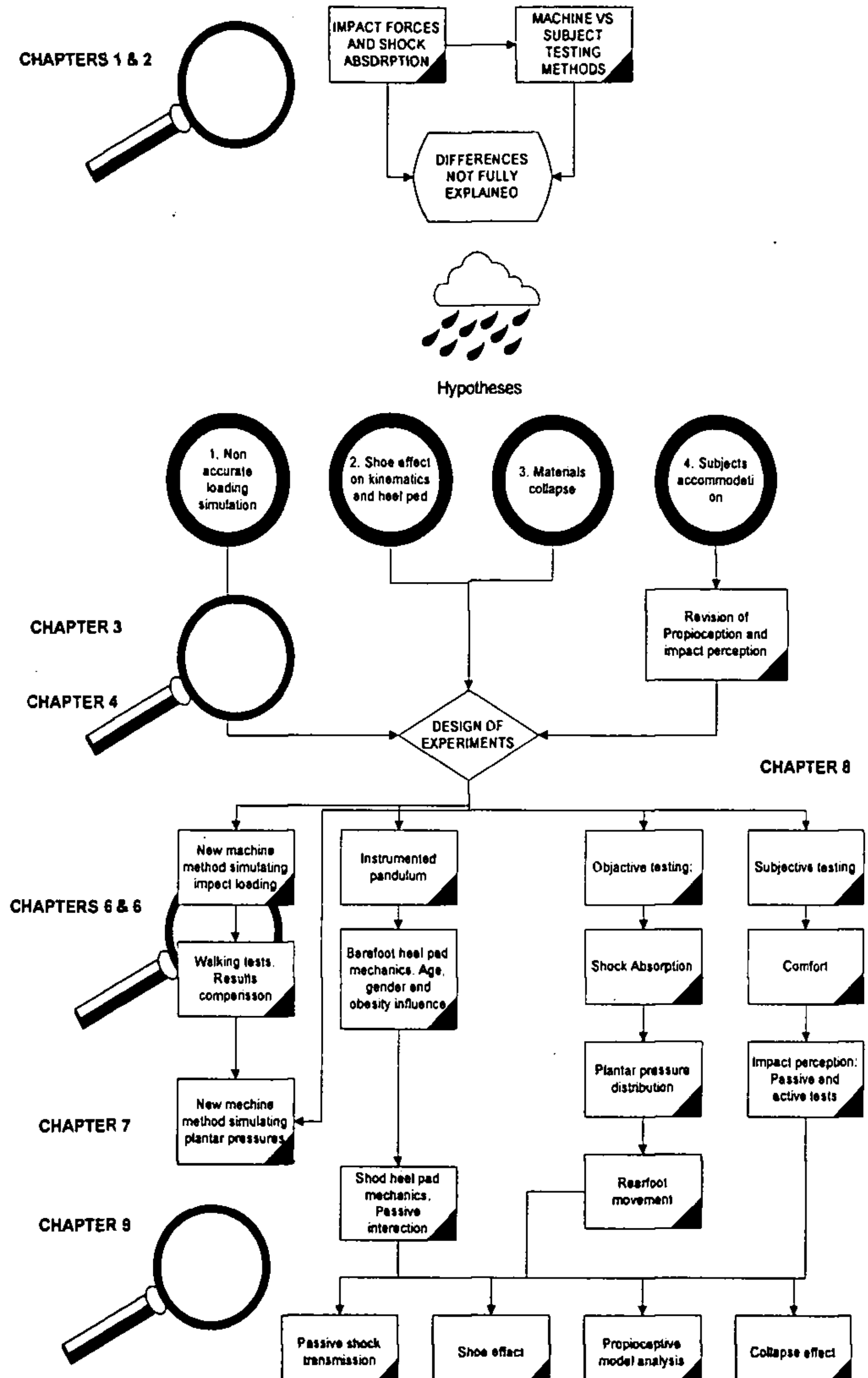

Figure 1.1. Thesis structure

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Introduction to absorption and attenuation of impact forces during walking

2. Introduction to absorption and attenuation of impact forces during walking

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The current understanding and underpinning knowledge, of shock absorption during walking and related terminology are included in this Chapter. The definitions and concepts presented are considered necessary for the understanding of this thesis.

2.1. Impact forces during human walking

An impact is a force of high magnitude applied for a short period of time (Robbins and Gouw, 1990). During walking the first foot-ground contact occurs at the heel (known as heel strike) and the resultant impact force initiates a transient stress wave that then travels through the musculoskeletal system to the head. The impact manifests itself as an initial peak ground reaction force when registered with force plates (Ground Reaction Forces, Figure 2.1) (Chu & Yazdani-Ardakani, 1986; Collins & Whittle, 1989; Johnson, 1986; Jorgensen & Bojsen-Moeller, 1989a; Munro et al., 1975; Simon et al., 1981).

This peak ground reaction (impact) force occurs during the initial 50 ms of the support phase of the walking cycle (when the foot is on the ground) with a duration from 5 to 25 ms (Folman et al.; Jefferson et al., 1990; Jorgensen & Bojsen-Moeller, 1989a; Ligth et al., 1980; Shorten & Winslow, 1992). The magnitude of these forces ranges between 0.5 and 1.25 times the body weight (Jefferson et al., 1990; Lewis et al., 1991).

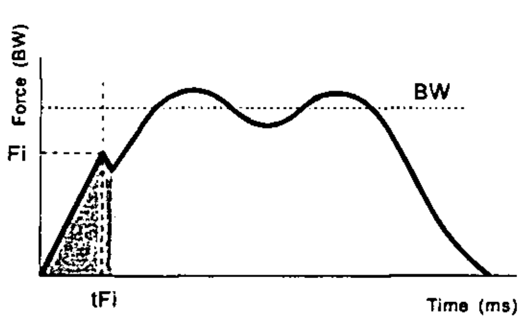

Figure 2.1: Ground Reaction Forces during gait.

The magnitude and duration of these impact forces depend on factors such as walking velocity (Clarke et al, 1985; Shorten & Winslow, 1992,), walking surface (Hamill et al., 1984), footwear (Light et al., 1980, Shorten & Winslow,. 1992), body weight (BW) (Shorten & Winslow, 1992) and human characteristics such as gender, age or obesity (Bates et al., 1983).

For shod walking and an impact of 15 ms duration, impact forces are typically 0.5 BW whilst barefoot walking registers higher forces and shorter response times (Folman et al., 1986). According to Collins & Whittle (1989), these values for heel strike duration and impact forces meet the criteria for impulsive loading.

During normal walking, the lower limb is subjected to repeated heel strikes and cyclic impulsive loads with impulses up to 100 Hz (Alexander & Vernon, 1975; Collins & Whittle, 1989; Folman et al., 1986; Jefferson et al., 1990; Johnson, 1986; Munro et al., 1975; Paul et al, 1978; Perry, 1983; Simon et al., 1981; Voloshin & Wosk, 1981).

2.1.1. Transmission of heel strike impacts

The impact of the foot with the ground produces a sudden deceleration of the leg known as shock. This is a transient condition when the equilibrium of a system is disrupted by a suddenly directional change of force. That causes an acceleration-deceleration wave known as a shock wave (a spatial propagation of mechanical discontinuity of a system (Nigg et al., 1995b)) that propagates up the musculoskeletal system to the head. (Light et al., 1980; Voloshin & Wosk, 1981, 1982, Wosk & Voloshin, 1981, 1985; Chu & Yazdani-Ardakani, 1986; Johnson, 1986; Collins & Whittle, 1989; Jefferson et al., 1990; Lewis et al., 1991; Shorten & Winslow, 1992; Noe et al., 1993). Such transmission shock waves have been recorded at accelerations levels of between 2 g and 15 g by accelerometers placed at both the distal and proximal ends of the tibia (Figure 2.2), and in the femur, sacrum, cervical spine and forehead (Alcántara et al., 1996; Folman et al., 1986; ; Forner et al., 1995; Hennig & Lafortune, 1991; Lewis et al., 1991; Light et al, 1980; Nigg and Herzog, 1994; Shorten & Winslow, 1992; Voloshin, 1993a). Some authors (Bromage et al., 1985; Buzo Cordova, 1983) have shown that the vertical component of the transmitted stress wave is accompanied by transverse components of the similar magnitude.

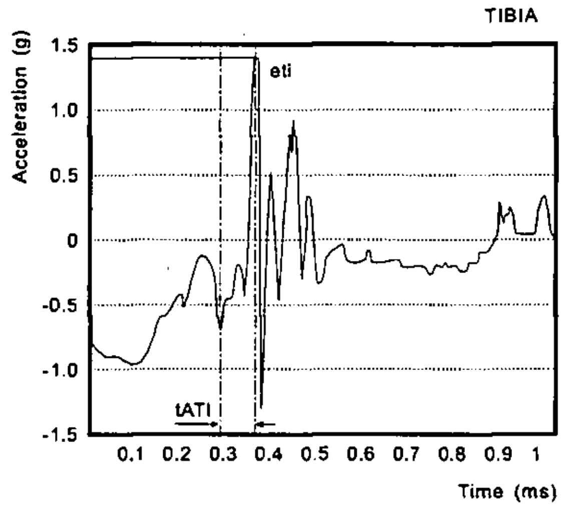

Figure 2.2. Acceleration signal registered at the tibie.

2.1.2. Relevance of heel strikes

Impact forces of three times body weight and accelerations to 17 g (gravity units) have been registered during running (Table 2.1). Impact forces and their transmission have been associated with several pathologies and with bone growth and human comfort and these effects have been observed in adults, children and in elderly people. As a consequence, biomedical sciences such as orthopaedics, footwear design and rehabilitation have increasingly focused on the effects of heel strike impacts.

Experiments with animals (Johnson, 1986; Radin et al., 1978, 1982; Voloshin & wosk, 1982; Wosk & Voloshin, 1981) have correlated repetitive impulsive loading with the aetiology and progression of several pathological conditions, especially joint degenerative diseases.

Table 2.1. Typical values for impact forces end accelerations

	Velocity (ms^{-1})	Fmax/ BW	Frequency (Hz)	Tibia acel. (g)	Haad acal. (g)
Walking	1.3	0.27 - 0.55	1st peak: 1,6 Hz 2nd peak: 1,17 Hz	2-5	1
Heel-toe running	3 - 6	1.6 - 3.0	1st peak: 17 Hz 2nd peak: 5 Hz	5-17	1-3
Toe running	4	0,4	5 Hz	5-12	1-3
Jumping take-off	2 - 4 6 - 8	1.4 - 3.3 5.4 - 8.3	1st peak: 20 Hz 2° peak: 4.5 Hz		

Several studies with humans have correlated impulsive loading with damage to soft tissue, bones, lower limb, spinal joints and with degradation of biomechanical properties of the articular cartilage and subchondral bone in the knee (Figure 2.3). This causes a variety of common diseases such as chronic low back pain (LBP), plantar fasciitis, occupational headaches, Achilles tendonitis, cartilage fracture, muscle problems and, specially, stress fractures and osteoarthritis (Bierlig- Sorensen et al., 1990; Gebauer et al.; 1987; Hoshino & Wallace, 1987; Jefferson et al., 1990; Johnson, 1986; Jorgensen, 1985; Jorgensen & Ekstrand, 1988, Jorgensen & Bojsen-Moeller, 1989a; Jorgensen, 1990a; Lewis et al., 1991; Light et al., 1980; Milgrom et al., 1985; Nigg, 1986b; Paul et al.; 1978; Radin & Paul, 1971; Radin et al., 1973; 1982; Shorten & Winslow, 1992; Simon et al., 1972; Voloshin & Wosk, 1981; 1982; Voloshin, 1993b). The incidence of these diseases is very high. Ostheoarthritis, for example, afflicts over the 50% of the population (Huskisson, 1979) and 85% of persons aged 70-79 years (Howell et al., 1978). Low back pain, a common medical disorder, is likely to afflict 90% of the population at least once in their life (Collins & Whittle, 1989).

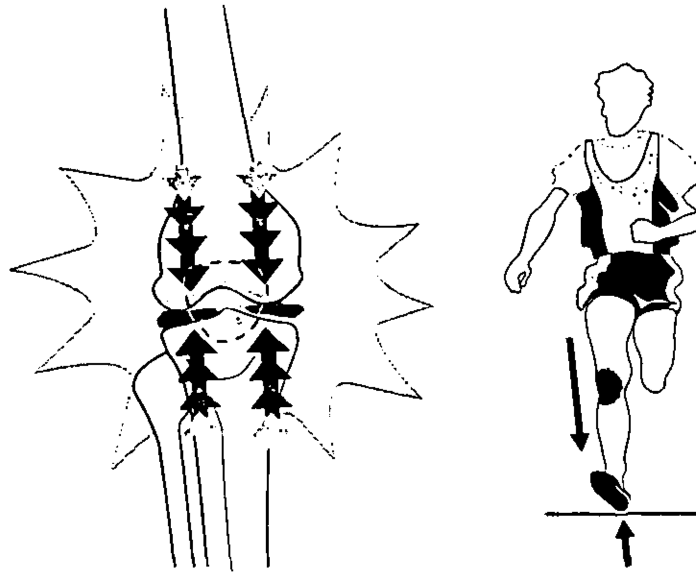

Figura 2.3. Heel strika impacts have been corralated with joint degenerative disaases

Impact forces have also been related to general negative haematological conditions such as increased red blood cell breakdown, observed in long distance runners (Falsetti et al., 1983; Shorten & Winslow, 1992), and to the loosening of hip prostheses with time (Hoshino & Wallace, 1987; Light et al., 1980). In contrast, experiments using animals have shown that repetitive impulsive loading is an important factor in bone growth and repair (Jankovich, 1972; Voloshin & Wosk, 1981). Hunter et al. (1945) related low impact forces to the loss of bone density. More recent studies (Fehling et al., 1995; Heinonen et al., 1995) of impact loading and bone mineral density (BMD) have reported a greater BMD increase in the legs and pelvis in female athletes in high impact sports like gymnastics or volleyball than in low impact sports such as skating. These findings are important and should be taken into consideration when assessing impact forces as stimuli for child growth (Alcántara et al., 1996).

Recent work has also correlated the level of impact forces with comfort during walking. Van Jaasveld et al. (1990) related the level of transmitted shock waves with comfort of people wearing below-knee prostheses. Concurrent preliminary studies at the Institute of Biomechanics of Valencia (IBV) concluded that there is a significant relationship between perceived excessively hard shoe sole and low levels of footwear comfort. Another study established a relationship between an excessively hard sole and increased pain in the heel and in the lumbar area (García et al., 1994).

2.2. Shock Absorption and Attenuation

As already stated, impact forces and stress waves can have very different effects, from the damaging to the beneficial, depending on their magnitude, duration and on individual's physical characteristics. Very high impact forces can be damaging whilst very low impact forces can lead to a decrease of bone mineral density and a loss of bone growth stimuli.

The first studies on shock absorption focused on running shoes because high and repetitive impacts occur during running. Initially the overall aim was to reduce as much as possible the impact forces and to attenuate the resultant stress waves. More recently, however, this subject has become controversial and there is little consensus on the relationship between magnitude and duration of impacts and on their effects. Whilst it may be possible to postulate an optimum level of shock absorption and attenuation to avoid damage and retain the positive effects such as growth stimulus, this optimum is dependent on the characteristics of locomotion and the anatomy and physiology of the human body. The management of heel strikes, therefore, should focus on controlling rather than reducing the effects and this is best achieved by means of shock absorption. However, current knowledge is incomplete and a deeper understanding of this subject is essential.

According to several authors (Jorgensen & Bojsen-Moeller, 1989a; Nigg & Cole, 1995) **shock absorption** is the ability of a given material, system or mechanism to decrease the effect of impact forces by means of energy absorption and dissipation. **Shock attenuation** is defined as the reduction of the shock wave transmitted after heel strike and is usually expressed as the ratio given by the acceleration registered at different parts of the body (e.g. the head) divided by femoral acceleration (Nigg & Herzog, 1994; Voloshin & Wosk, 1981). Other authors (Light et al., 1980; Maclellan & Vyvyan, 1982; Nigg, 1986b Radin, 1987 Voloshin & Wosk, 1981) have regarded shock absorption as a simple way to reduce loading in the human body and have used it as a palliative to protect the musculoskeletal system from damage due to impact forces. However, the effects of heel strike depend much on the shock wave transmitted through the body and this demands further investigation.

Some confusion about terminology in the literature was evident. Systems used for reducing heel strike loading are referred to as shock absorbing systems and methods for assessing their effectiveness or efficiency usually measure parameters related to both impact forces and shock transmission. Therefore, in this report when discussing the literature, **shock absorption** and **shock absorbing** systems are used in the generic sense described above. It is important to understand that these are different concepts though: lower impact forces (i.e. shock absorption) does not necessarily mean lower shock transmission (i.e. shock attenuation) since in some cases it is possible to reduce impact forces and increase transmissions as a result.

Shock absorbing systems (for shock absorption and attenuation) may be divided into **Natural systems** and **External systems** (Lewis et al., 1991; Yingling et al., 1996). The human body has inherent natural shock absorption systems that absorb loads arising from walking on naturally occurring surfaces. However, when the loads surpass physiological levels, the capability of these systems is exceeded and musculoskeletal damage often results (Jorgensen & Bojsen-Moeller, 1989[a]; Voloshin & Wosk, 1981, 1982; Wosk & Voloshin, 1981). When this happens it is necessary to supplement natural systems, either by improving their efficiency or by adding external shock absorption systems such as orthoses, special footwear, viscoelastic materials, etc. (Jorgensen & Bojsen-Moeller, 1989a; Milgrom et al., 1985; Voloshin & Wosk, 1982; Voloshin, 1993b).

2.2.1. Natural systems

Natural systems are the body's inherent mechanisms that absorb and attenuate impacts (Chu & Yazdani-Ardakani, 1986; Noe et al., 1993; Voloshin & Wosk, 1981, 1982). In 1973 Radin et al. showed that naturally occurring musculoskeletal shock absorption (SA) appeared to protect joints from potentially deleterious impulsive loading.

The role of the body's natural systems has been widely evaluated and, although results differ slightly, authors agree its contribution to shock absorption is very important. Natural systems are known to have attenuated 50% to 90% of the shock wave when it reaches the knee (Cavanagh & Lafortune, 1984; Forner et al., 1995; Light et al. 1980; Noe et al, 1993; Wosk & Voloshin, 1981) and up to 98% when it reaches the head (Forner et al., 1995; Light et al. 1980; Shorten & Winslow, 1992; Voloshin & Wosk, 1981; Wosk & Voloshin, 1981; 1985). Any external shock absorption facility, therefore, will benefit people with impaired natural shock absorption capability, due perhaps to injury or age, or if exposed to severe impulsive loading conditions because, for example, they are overweight, carry heavy loads or wear restrictive footwear.

The mechanisms of natural systems may be either **active** or **passive** (Lewis et al., 1991). *Active systems* include neuromuscular activity, joint mobility, proprioceptive compensation and muscular tone. Movement and flexibility of joints such as the knee and ankle, subtalar and hip have important roles to play in shock absorption (Chu & Yazdani-Ardakani, 1986; Dyshko & Farbe, 1993; Lewis et al., 1991; Noe et al., 1993).

Passive systems are a function of the mechanical properties of the components of the musculoskeletal system such as bone, cartilage and soft tissues. Soft tissues and fluids in joints including viscoelastic structural deformation have been shown to contribute to shock absorption (Voloshin & Wosk, 1981, 1982). Other authors have shown a limited contribution to shock absorbing properties of intervertebral discus, meniscus, sinovial liquid, subchrondal bone, ligaments and tendons, cartilage and periarticular tissues in joints, as well as skin and subcutaneous tissues (Chu & Yazdani-Ardakani, 1986; Hoshino & Wallace, 1987; Lewis et al., 1991; Noe et al., 1993; Robinovitch et al., 1995; Shorten & Winslow, 1992; Smeathers, 1989; Voloshin & Wosk, 1982).

There is some controversy about the real contribution of both the active and passive systems to shock absorption, particularly at the joints where it is very difficult to separate the contribution of joint mobility and joint structural elements. Paul et al. (1978) attributed to passive systems a more effective attenuation of peak forces at higher frequency ranges (100 Hz), while active systems will appear to be more effective at lower frequencies due to response time of muscular reaction. The authors (Collins & Whittle, 1989; Pratt & Sanghera, 1989) consider that passive elements play only a small role in body protection from impulsive loading due to the viscoelastic nature of these systems that cause them to stiffen and reduce shock absorption ability at high frequencies under dynamic loading.

However, other authors (MacMahon et al., 1987 Nigg, 1986b; Simon et al., 1981) suggest that active attenuation systems cannot play a significant role because the duration of impact (20 ms) is less than muscle reaction time (70 ms). But several authors (Clarke, et al., 1983a) argue that muscles prepare for attenuating impact before it occurs (pre-activation) and thus compensate for the short reaction time.

A more recently proposition is that passive and active systems are related to mechanisms of accommodation under impact conditions. This proposition was investigated as part of this research.

A schematic representation (Figure 2.4) of shock absorbing systems in the lower leg shows knee and ankle movement as natural active systems, the heel pad as a natural passive system and underfoot materials as an external system.

2.2.1.1. Heel pad

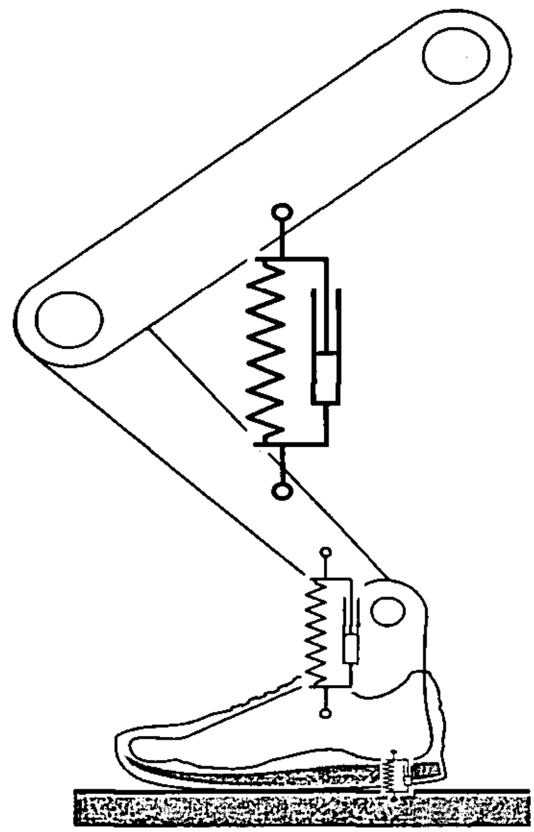

Figure 2.4. Schematic representation of main SA systems in the lower limb.

The heel pad is a flexible mass approximately 18 mm thick located between the calcaneus and the foot plant skin (Figure 2.5) and has been described as the most important shock absorbing system of the foot and ankle complex. It can absorb up to 90% of impact energy (Chu & Yazdani-Ardakani, 1986, Jorgensen & Bojsen-Moeller, 1989a; Noe et al., 1993).

The heel pad is a unique structure with a fat content that is less saturated than anywhere else in the body (Winter et al., 1988). Nature recognised the importance of the fat cells in heel pad in times of famine as these cells were spared from nutritional metabolism (Batty Shaw, 1902; Wells, 1940).

The passive natural shock absorbing system has attracted the interest of many researchers due to its special mechanical properties under impact loading. Understanding the normal function of the heel pad is of great importance to practitioners in the areas of gait analysis sports medicine and athletics coaching. Knowledge of the mechanical properties of the heel pad is helpful when designing effective walking and running footwear.

Again, there has been some controversy concerning the real shock absorbing ability of the heel pad. Some authors argue that the heel pad bottoms-out easily under relatively low compressive loading and serves only as protection of the calcaneus without any significant role in shock absorption during walking and running (Aerts et al., 1993a; Aerts & De Clercq, 1993c; Robbins & Gouw, 1989;). However, in many of these reports the methodology described was suspect in both experimental set up and measuring techniques (Chapter 7). Nowadays the heel pad is widely accepted as an important shock absorber although there is no general agreement on its actual properties. It is now universally accepted that experimentally obtained measurements of shock absorption are due to the inter-action between the different systems and not due to the heel pad alone and that the influence of leg and body movement is significant (Aerts et al., 1995; Cavanagh et al., 1984b; Kinoshita et al., 1996a; Light et al, 1980; Paul et al., 1978).

The alternative to machine testing is *in-vitro* experiments. The accepted disadvantage of *in-vitro* experiments is the inability to isolate the heel pad from the other systems of the lower leg (ankle, knee, etc.). This is not a factor in this research programme since its objective is to investigate lower leg shock attenuation. However for convenience and consistency with previous work, the mechanical properties of the lower leg are referred to as those of the heel pad.

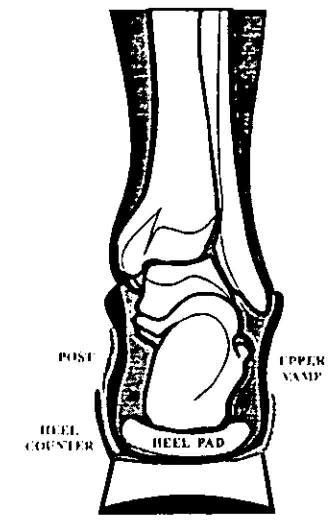

Figure 2.5. Anatomy of the hindfoot showing the heel pad.

The shock absorbing properties of the heel pad are closely related to its highly specialised physiological structure (Jahss et al., 1992a; Jorgensen & Bojsen-Moeller, 1989a). The first known study on the anatomy of the heel pad (Tietze, 1921) described the fat pad as a thick absorbing structure made of several adipose tissue-filled compartments securely tied by connective tissue bands.

Blechschmidt (1933) produced a more exhaustive work and his observations revealed denser septa on the lateral side than on the medial side. This could be regarded as an adaptation to human locomotion given that initial contact of the foot with the ground usually takes place in the lateral aspect of the heel.

Recent works have presented complete and detailed anatomical description of the heel pad as a complex structure with neural, vascular, fibrous and elastic components intertwined with fat cells. (Cavanagh et al., 1984b; Jahss et al., 1992a; 1992b; Jorgensen & Bojsen-Moeller, 1989a; Khuns, 1949; Prichasuk et al., 1994a; Prichasuck, 1994b; Radin & Paul, 1971; Radin et al., 1973; Voloshin & Wosk, 1981, Wallace, 1982;).

The normal structure of the heel pad consists of fat cells enclosed in a fibroelastic structure formed by dense bundles of elastic tissue (mainly collagen and elastine), anchored to the bone and the dermis. They are U-shaped or comma-shaped fat-filled columns arrayed in a basically vertical orientation, a honey-combed configuration without intercelular space. All regions of fat cells are totally enclosed by septae, forming an overall mechanical closed-cell structure.

It has been suggested that the honey-comb closed cell structure of the fat pad serves to provide the mechanical integrity for shock-absorption and that its mechanical properties depend on the integrity of the septae of every individual cell (Jahss et at., 1992b; Prichasuck et al., 1994a). This is consistent with the observations that degenerated fat pads tend to have an abnormal septal pattern and that load bearing capability of the heel pad is related with its integrity (Jahss et al., 1992b).

The structure of the septae is designed to withstand compressive loads. It consists of kinked fibro-elastic bundles that stiffen and absorb energy under compression as the fibres are compressed and straighten as a part of the resistant structure. This mechanism is not unlike the structure of some rubbers in which the disordered bundles of molecules absorb energy under load by molecular reorganisation.

Similarly, since the fat core is incompressible, as the cell walls become thicker or more fibrous they become more difficult to deform under load. On the other hand, if the walls break down the structure becomes less resistant to compression and deform–more easily thus reducing the mechanical properties of the heel pad. This will cause the heel pad to bottom out and reduce its recovery-ability when the load is removed thus determining the shock absorbing capability of the heel pad.

Under vertical loading, the structure in the central heel pad is prevented from severe bulging by internal reinforcement structures. The septae are reinforced internally with elastic transverse and diagonal fibres that connect the thicker walls and separate the fat into compartments or cells. Their upper part is fixed to bone or other septae and the lower part to other septae or to a thick, fibrous, subdermal layer termed the "internal heel cup". In this context Kinoshita et al. (1993a) reported that insufficient heel pad function capacity can lead to the development of shock-dependent injuries, such as heel pain (Khuns, 1949), plantar fasciitis (Sewell et al., 1980) and Achilles Tendonitis (Jorgensen, 1985).

The mechanical behaviour of the heel pad under impact loading is non-linear (force - displacement relationship depends on loading level) and viscoelastic (displacement depends on loading frequency) as shown by the force-displacement curve (Figure 2.6). This curve shows a region of low stiffness followed by a region of high stiffness and displacement continuing to increase after maximum force has been reached. At zero force a residual deformation is observed which will recover on the time (Cavanagh & Lafortune, 1984a; Valiant, 1984).

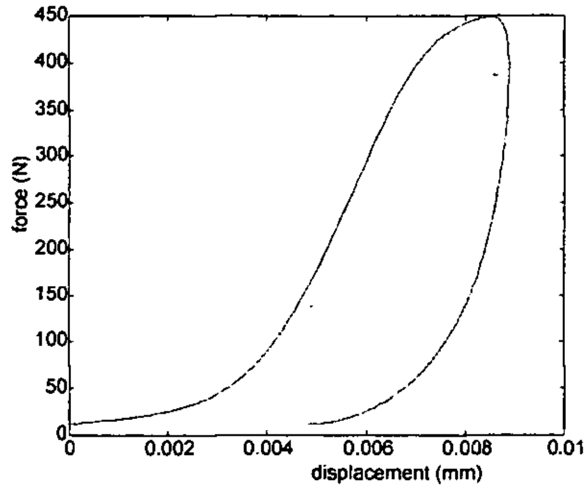

Figura 2.6. Force-displacement curve of a normal heel pad obtainad in this thesis using a ballistic pandulum (v = 0.62 m/s).

The knowledge and understanding of heel pad properties is very important for pure and applied biomechanics, but results of studies on heel pad properties are not always consistent. A great variety of methods and parameters have been used to describe the heel pad behaviour under impact loading making the comparison of results of different authors very difficult. The relationship between the various parameters and their relationship with impact force reduction describing heel pad mechanics has not been show.

On the other hand, the literature refers mainly to running conditions providing scarce data that could be applied when dealing with shock absorption during walking. It has to be considered that shock peak forces reach lower levels and reduced time response during walking. Thus, behaviour of non-linear and viscoelastic materials such as the heel pad will be different when walking and when running.

The literature search confirms that the shock absorbing properties of the heel pad vary greatly from person to person (Jorgensen et al., 1989b) and heel pad properties may depend on personal characteristics (Jahss et al., 1992b; Jorgensen et al., 1989b; Kinoshita, et al., 1996a; Prichasuck et al., 1994a; Prichasuck, 1994b). But a clear relationship with personal characteristics such as body weight, physical history, gender and others not yet established (Jorgensen et al., 1989b). There are some studies on isolated human characteristics such as age (Kinoshita et al., 1996a) but the multi-factor studies had small and inconsistent samples (Jorgensen et al., 1989b). The important concepts of repeatability and reliability have been frequently overlooked.

Thus, in this research, understanding the influence of human characteristics in heel pad properties assumed a new importance. The first properties to be measured tended to characterise mechanical integrity under compression with respect to heel pad thickness and compressibility. As that work progressed, shock absorbing ability and associated parameters became the main focus of the research and this, in turn, led to the development of new test methods. Some heel pad characteristics obtained from the literature review are detailed below.

Thickness was the first heel pad property measured. Thickness has been widely used for heel pad characterisation in clinical assessment. It has been correlated with the elastic properties of the heel pad to show that as the pad thickens it becomes stiffer and less absorbent. A thicker heel pad has been associated with plantar fasciitis and plantar heel pain (Amis et al., 1988; Prichasuck et al., 1994a; Prichasuck, 1994b) and a very thin heel pad with foot ulceration in diabetes (Gooding et al., 1986). Heel pad thickness in normal people range from 12.5 mm to 30 mm with a mean value of 18 mm. Males have a greater thickness of heel pad and it tends to increase with body weight and age (Jorgensen et al., 1989b; Prichasuck et al., 1994a; Prichasuck, 1994b).

Different methods have been used for measuring thickness including X-ray to ultrasound (Gooding et al., 1986; Jorgensen, 1985; Jorgensen et al., 1989b; Khuns, 1949; Steinbach & Russell, 1964).

Visual compressibility index has been widely used in clinical assessment. Jorgensen (1985) defined this index as the ratio between the thickness of the heel pad when unloaded and the thickness loaded as measured by lateral x-rays (Figure 2.7). The index is 14/25 for a normal heel pad and 57/45 for a pathological heel pad (Jorgensen, 1985) and, in general, increases with age and body weight. The smaller this ratio the less elastic the heel pad. In this context, Jorgensen (1985) showed a correlation of the compressibility of the fat pad with Achillodynia (a pathology of Achilles tendon) and Prichasuck et al. (1994a, b) found that the compressibility index was greater in people suffering from plantar heel pain

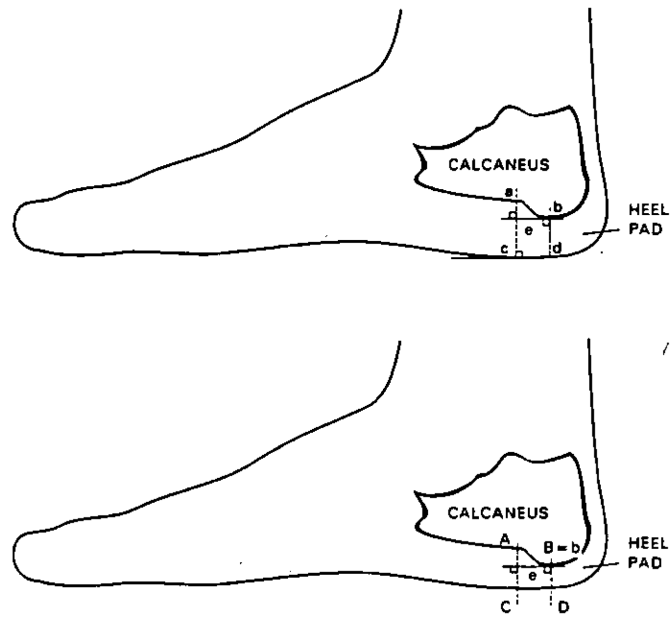

Figure 2.7. Definition of the compressibility index. CI = (B-D/b-d)/(A-C/a-c). Upper: unloeded; lower: loaded

The stiffness of a material is its resistance to deformation under load. The heel pad is viscoelastic and non-linear and thus stiffness depends on loading magnitude and frequency and any study of the heel pad is complicated as it has no single stiffness parameter. Different methods to measuring stiffness have been used in the study of the heel pad. The greater differences are found between *in-vivo* and *in-vitro* testing. The stiffness value when calculated from the load-displacement curve (Figure 2.6) depends on whether it is measured during the initial or final loading phase. Some authors used the maximum stiffness which occurs in the mid region of the loading curve, others the stiffness at peak force (Valiant, 1984) and stiffness at a loads similar to body weight has been also computed (Aerts et al., 1995). Values reported in the literature range from 10E6 to 10E5 N/m. Valiant measured an initial stiffness of 8000 N/m and a final stiffness of 105646 N/m (Cavanagh & Lafortune, 1984a; Denoth, 1981; Jorgensen et al., 1989b; Misevich & Cavanagh, 1984; Nigg & Denoth, 1980;).

Stress-strain curves for the heel pad are very difficult to obtain due to the complex and rapidly varying contact area of the heel pad when loaded. Nevertheless, some approaches are to be found in literature.

Nigg and Denoth (1980), in a human drop test, presented an approach to calculate stress-strain curves of the heel pad under impact loading using a planar heel during heel strike. He described stress-strain curves for tests with 10 participants. However, he did not describe his methodology nor, in particular, details of the procedure used to calculate stress in the heel pad, a compressible material. For stresses from 5 to 15 $kNmm^{-2}$ deformations between 2.5 and 8 mm were reported. Great individual variability was found.

The reduction of impact forces is related to *shock absorption.* This ability of the heel pad has been observed both in walking and running human tests and in machine testing. According to the ISO 4651, the drop test standard, shock absorption is evaluated as the reduction of impact force with respect to the maximum impact force registered.

In walking tests, the reduction of impact forces with shoes relative to barefoot (usually the highest impact condition) is used for the evaluation of footwear (Jorgensen & Ekstrand, 1988). Generally, the greater the force reduction the more effective the footwear and, by this definition, the heel pad has been observed to be an effective shock absorbing system (Jorgensen & Ekstrand, 1988; Jorgensen et al., 1989b).

***Shock attenuation* refers to he reduction of shock transmission through the body after heel strike.** This has been studied both in temporal and in frequency domain. Some authors have suggested that measuring acceleration must be accompanied by kinematic analysis of walking for the results to be meaningful.
The reduction of shock transmission (shock attenuation) has been also widely used for characterising materials in walking and running experiments. Shock transmission is defined by the relationship between impact forces and accelerations registered at different locations of the body, and between acceleration peaks (Alcántara et al., 1996; Forner et al., 1995; Light et al., 1980; Nigg & Denoth, 1980; Voloshin & Wosk, 1981, 1982). The reduction in magnitude of the acceleration peaks at different locations in the body may be used to analyse shock attenuation as the ratio between acceleration at different body locations. Ratio between acceleration in tibia and impact force is so defined as tibial shock transmission whereas the quotient between head and tibia acceleration as head transmission (Alcántara et al., 1996; Forner et al 1995). In general, the lower the transmission the better the footwear material and the greater the attenuation.
Noe et al. (1993), using an impact hammer *in-vitro* and a mechanical model, observed an attenuation of peak acceleration of 80% for the human heel pad, whereas for EVA (Ethyl Vinyl Acetate) it was only 55%.
The **Shock Factor** (Johnson, 1988) has been defined to study the effectiveness of shock absorbing materials in the frequency domain. The severity of stress wave is determined by the amplitude and frequency. So, the effectiveness of a shock absorbing material may be measured by its ability to reduce the higher frequency components and the magnitude of the stress wave. The shock factor is defined as the ratio *As/(As+An)* where *As* is the area under the curve between 50 and150 Hertz and *An* the area between 10 Hz and 50 Hz. Fifty Hz is the frequency observed in force

plate measurements which distinguishes between forces corresponding to normal walking and those due to the impact. The higher the shock factor the lower the shock attenuation.

The ***energy absorption*** of a material is related to the reduction of both impact forces and shock transmission. Important differences are found in the literature depending on the methods used for characterising the energy absorption of the heel pad. Absorption values obtained ranged from 70% to 95% of input energy *in-vitro* and between 30% and 50% *in-vitro* for running loading (Aerts et al., 1995; Bennet & Ker, 1990; Cavanagh & Lafortune, 1984; Nigg & Denoth, 1980; Valiant, 1984). Kinoshita et al. (1993a, b; 1996a) using a drop test method measured values of about 76% (72-84%) for a velocity impact of 0.94 ms-1 and 77% for a slower velocity impact of 0.58 ms-1, whilst values registered using pendulum test method are usually 10% lower.
From a mechanical point of view, when a material is deformed, the energy absorbed is related to the elastic, plastic/viscous and viscoelastic behaviour of the material. Both elastic and viscoelastic deformation return to original thickness after loading, but while elastic materials return 100% of loading energy viscoelastic materials return only part of it, dissipating the remainder. Plastic materials show a residual deformation at zero force with energy dissipation due to plastic deformation energy. Dissipation by heat and other mechanisms is observed and a typical hysteresis loop in the force-displacement curve is obtained (Figure 2.6).
Energy absorbing materials may exhibit elastic, viscous and viscoelastic behaviour depending on the speed of impact and its duration. The heel pad behaviour under impact loading seems to be a complex interaction of elastic and viscoelastic phenomena, but this has not been clearly established.

2.2.1.2. Knee

The shock absorption provided by the knee is mainly due to muscular control and to viscoelastic deformation of structural elements. The former will mainly reduce impact forces (shock absorbing role) whereas the latter will attenuate shock transmission (shock attenuation).
Shock absorption by muscular activity is mainly associated with its role in the control of joint mobility. This was recognised as important as early as 1960 when Hill (1960) described the contribution of active joint muscular contraction in energy absorption. The contribution of joint mobility to shock absorption is associated with eccentric muscular action (with change in muscle length) (Voloshin, 1982).

Hoshino & Wallace (1987) attributed an important role to muscular activity in modifying impulsive loading in the knee during walking, running and jumping. Jefferson et al. (1990) showed that quadriceps' activity prior to heel strike is very important in controlling foot landing, thus influencing the impact forces. Electromyography (EMG) measurements (Inmann et al., 1981) show that activity in quadriceps and hamstrings prior to heel strike decelerated the leg at the end of the oscillation phase of the walking cycle (before the foot is in contact with the ground) by controlling the knee, which finally slows down the fall of the leg.

Radin et al. (1991) studied the role of neuromuscular control of leg movements in preserving the knee from impact loading and showed significant differences between a normal-knee group and a painful-knee group. The time to heel-strike was lower in painful-knee group which gave less time to prepare for the impact with subsequent greater angular velocity of the tibia and downwards velocity of the ankle at heel-strike and greater impact forces. The painful-knee group seemed unable to decelerate the leg before heel-strike through muscular activity and this increased the impulsive loading. For this group, peak accelerations in tibia and femur were greater with a greater loading rate suggesting an impulsive loading of greater magnitude and frequency making the painful-knee group participants prone to develop osteoarthritis.

Although muscular action is very important in shock absorption in the knee, the contribution of the different passive elements also has been widely studied. The intra-articular structures of a normal knee such as meniscus, cartilage and compact bone are effective absorbers of shock stress waves in the leg. Loss or damage of these structures will result in a damaging level of shock transmission that could lead to the development of degenerative arthritis in other joints. Pain or implanting prosthesis, in the knee means a reduction of natural shock absorption and care should be taken to avoid attendant negative effect in hip and spine.

The contribution of the different elements of the knee has been assessed through *in-vitro* studies. Periarticular soft tissues and subchondral bone seem to be the most effective systems (Chu & Yazdani-Ardakani, 1986; Hoshino & Wallace, 1987; Radin et al., 1978, 1982). Other works (Hoshino & Wallace, 1987) showe that removing the meniscus increased impact forces.

However, the role of cartilage is less clear. Some workers (Radin et al, 1973) showed no shock absorption in the cartilage, however *in-vitro* studies concluded that after the stiffening of other structures cartilage degeneration leading to osteoarthritis. It appears that the shock absorbing ability of the knee diminishes if the cartilage degenerates but its removal, however, has little effect on to shock absorption (Chu & Yazdani-Ardakani, 1986; Radin et al., 1973; 1978; 1991; Simon et al., 1972).

2.2.1.3. Spine

The relevance of the spine in shock absorption is strongly supported by the relationship found between low back pain and heel strikes (Voloshin & Wosk, 1982). The transmission and attenuation of dynamic forces in spine rely mainly on the intervertebral discs (Voloshin & Wosk, 1982) which act as a shock absorbers attenuating the impact forces by compressive displacements (Smeathers, 1989). Gebauer et al. (1987) showed that due to stiffening at high frequencies, the discus could not attenuate as previously thought. Its natural frequency was measured to be in the range 45-70 Hz and since heel-strike impact range between 10-75 Hz, the discus was considered unable to attenuate impulsive loading with frequency components lower than 100 Hz. However, the intervertebral joints may form a more complex system that acts as flexible chain links which will reduce the resonance frequency of the spine that would allow the spine muscles and ligaments to dissipate energy. Impact forces of principal frequencies below 40 Hz were registered in *in-vivo* studies (Smeathers, 1989), using accelerometers mounted in the sacrum and lumbar back. The normal spine was able to attenuate this frequency (actually above 20 Hz.) but in an anquilosed spine this ability was not observed which could be due to a loss of flexibility.

Helliwell et al. (1989) presented similar results. He found that, whereas the intervertebral discus is able to attenuate peak forces above 100 Hz, a healthy spine attenuates loads above 15 Hz. This difference was attributed to the flexibility and bending in different planes of the spine supported by the fact that a spine unable to bend (due to ankylosing spondylitis) did not show this attenuation.

2.2.2. External systems

External systems or artificial shock absorbers include actions aimed at either ***improving or supplementing*** the shock absorption and attenuation provided by natural systems.

These systems should be used when impact loading exceeds the capability of natural systems and extra shock absorption is needed to avoid damage. This can occur with reduced shock absorption of natural systems or when impact loading is higher than usual.

The efficiency of natural systems is reduced by:

i.- Musculoskeletal disease or injury (Jorgensen, 1985; Jorgensen & Ekstrand, 1988; Noe et al., 1993; Voloshin & Wosk, 1981, 1982; Wosk & Voloshin, 1985).

Stiffening of locomotor system diminishes its shock absorbing properties. Jorgensen (1985) related a loss of shock absorbing properties of heel pad with higher impacts, which led to an increase of loading in the Achilles (Light et al., 1980, Voloshin & Wosk, 1981). On the other hand, degenerative diseases also diminish these properties in the affected joint.

ii.- Fatigue. Experimental evidence has shown that muscular fatigue influences shock absorption and attenuation. Voloshin (1993a) observed a reduction of 35% in shock absorption after 24 minutes running at 15 Km/h, whilst Light et al. (1980) showed that when fatigue appears muscle control becomes less efficient resulting in a decreased shock absorption. Verbitsky (1998) showed an increase of peak acceleration transmitted as fatigue occurred during running. In these circumstances, external absorption is important.

iii.- Negative influence of footwear in joint mobility caused, for example, by safety boots, high heel shoes, etc. (Voloshin & Loy, 1994).

On the other hand, impact loading reaches higher levels than usual due to:

i.- Overweight, due to weight increase, articular problems and differences in body fat content associated to obesity.

ii.- *Walking on hard surfaces or hard heels* which are common elements in daily activities. Less compliant materials are usually associated to greater impact forces (McMahon & Greene, 1979).

iii.- *Walking on high heels* affect joint absorption and foot position at initial contact (Voloshin & Loy, 1994) and results in greater peak forces and accelerations.

iv.- *Running and jumping* which increase body acceleration and inertia so that ground contact results in greater impact forces (Table 2.1), and

v.- Walking whilst carrying loads increases impact loading by both weight increase and modification of body posture.

Thus, as the efficiency of natural systems decreases a growing importance is given to external systems. Using an insert for extra shock absorption is almost essential for people with orthopaedic problems, people whose work involves walking for a long time and people standing for a long period (Lewis et al., 1991). But, even under physiological normal walking conditions, the repetitive and continuous action of stress waves arising at heel-strike tends to degrade the natural systems with the degeneration of the articular cartilage and osteoarthritis.

External systems can be divided into **Improvement and summation systems**. The former are aimed at improving the function of natural systems whilst the latter supplement natural performance.

Early works on biomechanics found differences in the impact forces with shod and unshod conditions but no differences between different footwear. More recent works found differences between hard and soft footwear, between different insole materials and, generally, much greater differences between shod and unshod. These findings implied that the role of the sole was not important and that heel pad confinement and the kinematic control due to footwear contributed to the increase in shock absorption. Thus giving greater importance to improvement systems (confinement and kinematics) than to the summation systems (sole and insole) (Lafortune & Hennig, 1992; Lees & Roberts, 1984; Lees, 1988). Summation systems usually consist of modifications on external factors such as walking surface, footwear sole or insole (Clarke et al., 1983a; Frederick et al., 1984; Jorgensen & Bojsen-Moeller, 1989aLight et al., 1980; Nigg, 1986). These factors influence natural systems and can result in lower natural shock absorption and increased instability for the foot more compliant produce great variety of problems. Thus it would be preferable to improve natural systems rather than supplement them.

2.2.2.1. Improvement Systems

These systems increase shock absorption by improving the actuation of natural systems and aim to restore rather than substitute them. The most important and the best known is the confinement of the heel pad. This system acts by restricting the calcaneal heel pad by means of a rigid counter in the shoe or using a heel cup either in the foot bed or in the insole. The confinement prevents the heel pad from bottoming-out thus enhancing its properties. It has been showed to be very effective with degenerated and injured heel pads (Bojsen-Moeller, 1989a; Jorgensen & Ekstrand, 1988; Jorgensen &) augmenting natural shock absorption up to about 50 % (Jorgesen & Bojsen-Moeller, 1989a). There are contradictory results, however, depending in the toughness of the restrain or confinement (Valiant, 1984; De Clercq et al., 1993; Jorgensen & Ekstrand, 1988;).

Other systems that are currently being studied are for ankle support without limiting its shock absorption, artificial cartilage (Oka et al., 1992) and the role of human accommodation to walking conditions in Shock Absorption. These systems are currently less evolved and there exists a general lack of work and analysis from a biomechanical point of view. But, recent results (Brizuela et al., 1997; Robbins and Gouw, 1990; 1991) present these subjects as alternative future solutions to summation systems. This thesis focuses on them.

2.2.2.2. Summation systems

Summation systems aim to increase shock absorption by supplementing the natural systems. They involve adding materials or systems with enhanced shock absorbing properties mainly as sole or as insole of the shoe. Apart from very specific midsole sports shoe systems such as air bubbles, gel, and others there appears to be an adequate selection of materials that can improve general shock absorption and these are used mainly for footwear manufacture and a great variety in clinical practice.

These materials are generally viscoelastic and non-linear. Recommended revisions of viscoelastic materials used in clinical practice and in footwear manufacturing are provided by Whittle (1997) and Rome (1990) respectively.

Many materials are viscoelastic to a greater or lesser extent and may be divided into seven categories (Whittle, 1997; 1999), namely the polyurethane elastomers (sorbothane, viscolas, etc.), polyurethane foams (Poron, PPT, etc.), polyethylene foams (evazote, pelite, plastazote, etc.), polyvinyl chloride foams (Implus), ethylene vinyl acetate (EVA), synthetic rubber foams (neoprene, noene, spenco), and silicone rubber. In the footwear industry, materials most often used for soling and insoles are rubber, PVC, EVA, PU, polyethilene, TR (technical rubber), leather, latex, cork and mixtures. The mechanical properties, in general, and shock absorption, in particular, of these materials depend on their chemical composition and internal structure of materials (Whittle, 1997; 1999).

For clinical and footwear applications one of the most important properties of viscoelastic materials is their ability to reduce the effect of impact forces. The magnitude of the force between the heel and the ground depends on two things: the momentum transferred and the time of transfer. The momentum change is equal to the product of force and time ($\int F\,dt$), so the same amount of momentum may be transferred by a large force in a shorter time or by a smaller force in a longer time. Accordingly, impact forces may be reduced by extending the duration of the impact and by absorbing energy. Any elastic materials, including viscoelastic ones, are able to reduce the impact forces by increasing the time taken to transfer the momentum. However, reducing the peak force in this way requires the presence of some thickness of compressible material. If the thickness of material is inadequate, when fully compressed the phenomenon of "bottoming-out" occurs abruptly increasing the stiffness of the material. Additionally, a great compression could cause instability during walking.

Viscoelastic materials are also able to reduce the effect of impact forces by absorbing energy. Purely elastic materials return most of the momentum to the foot whereas viscoelastic materials return only part as some energy is absorbed. If a totally elastic material was able to store and return all of the energy, the change in momentum would be *2mv* (twice mass times velocity) as impact momentum would change from *mv* downwards on impact to *mv* upwards on rebound or recovery. In contrast, with a viscous material capable of dissipating all of the energy (totally plastic impact), the change in momentum would be only *mv*, since rebound would not occur. Viscoelastic materials show different vertical ground reaction forces due to differences in amount of momentum exchanged (Whittle, 1997; 1999).

Pratt et al. (1986) reported a trend towards the prescription and supply of shock absorbing insoles based in the viscoelastic properties of materials not only for proven clinical/medical conditions but for sporting and leisure activities and for the non-specific heel or foot pain. This has resulted in many new products that have not been fully proven to be suitable for the purpose intended.

Studies based on accelerometry have indicated that shock transmission was reduced by the use of footwear. Several workers have shown that either shoes or insoles made of materials with good shock-absorbing properties reduce tibia shock and transmission of acceleration relative to barefoot (Bierlig-Sorensen et al., 1990; Johnson, 1986; Lafortune & Hennig, 1992; Light et al., 1980; Van Jaasveld et al., 1990; Voloshin & Wosk, 1981; Voloshin, 1993b). Force plates have, however, provided conflicting results regarding the cushioning properties of footwear during walking. Several workers repor higher initial peak vertical forces and faster loading rate barefoot than for shod walking due to the bottoming out of the heel pad (Lafortune & Hennig, 1992). However, no differences have been reported for different types of shoes, contrary to other authors who found differences within shoes but no differences compared with barefoot activities (Lafortune & Hennig, 1992; Van Jaasveld et al., 1990). However, some reports showed that different insert materials result in different shock absorption ratios (Forner et al., 1995).

Light et al., as early as in 1980 showed that footwear reduces impact forces and stress waves during walking, registering higher heel strikes for harder shoe heels. In general, a higher shock absorption is observed in sports footwear compared to leather shoes (Lafortune & Hennig, 1992; Smeathers, 1989; Van Jaasveld et al., 1990). In this sense, main role of shoes in impact attenuation was not clearly attributed to material properties and the effect of shoes in ankle movement and of heel pad confinement seemed to have great influence and these functions need to be clarified.

Laboratory testing indicates that most viscoelastic materials provide useful shock absorption under physiological conditions. Different studies have shown that materials used in footwear and in orthesis manufacturing have a high shock absorbing effectiveness that reduces both impact forces and the amplitude of shock waves arising at heel strike (Johnson, 1986; Pratt et al., 1990; Voloshin & Wosk, 1982;).

Materials such as noene, sorbothane, poron, implus, tercolite, latex and others are the most used materials for shock absorption, presenting quite different properties (Forner et al. 1995; García et al., 1994). Sorbothane has been shown to reduce heel pain due to achyllodinia (a degenerated painful heel pad). Voloshin et al. (1985) registered a reduction of 40% in impact forces using a rubber insert and showed that viscoelastic inserts could absorb more than 17% of impact energy and up to 30% in some cases.

Thus, viscoelastic inserts have proven to be efficient in palliating a variety of diseases such as low back pain (Voloshin & Wosk, 1981) and pain from knee arthritis (Radin et al., 1991; Voloshin, et al., 1982;). Further, viscoelastic inserts or insoles have been shown to reduce the incidence of stress fractures and pathologies directly related with vertical loading, for example, metatarsalgias - heel pain and arch pain (Milgrom et al., 1985, 1992). Johnson (1988) found a reduction up to 30% of acceleration peak with absorbing footwear. Voloshin and Wosk (1981) showed a reduction of impact forces and transmission of 42% with the inclusion of an arch support of shock absorbing material.

Clinical experience (Voloshin & Wosk, 1981; Wosk & Voloshin, 1985) has shown a relatively quick improvement of low back pain when treated with viscoelastic inserts placed at the heel of the insole. The inclusion of a shock absorbing material like EVA in footwear used by paraplegics has been shown to reduce the transmitted acceleration with increased comfort during walking, but the result depended on material thickness (a heel of 20 mm reduces 12% whilst a 10 mm insert reduces 26%), (Bierlig-Sorensen et al., 1990). Rooser et al. (1988), using accelerometry in walking experiments found that heel strike deceleration in tibia in rheumatoid patients with a knee arthroplasty was lower when wearing compliant (polyeurethane) shoes than with leather rubber heeled shoes irrespective of the material used for a heel insert. So, no extra benefit in shock attenuation was observed by introducing a viscoelastic insert into a soft shoe.

A shortcoming of viscoelastic materials is that their properties degrade with wear time and are dependent in temperature, humidity and loading frequency but their shock absorbing properties remain higher than in other materials in most of cases (Johnson; 1986; Kinoshita et al., 1996b; PFI, 1988; Pratt et al., 1986, 1990,). It is necessary, therefore, to carefully select materials with reference to ageing and frequency range of loads.

From the literature search, it is evident that impact forces and shock waves arising during walking are major subjects of study in many fields of biomechanics. Nature has managed these by endowing the human body with affective means to attenuation such as the heel pad. However, different diseases and modern life resulting in an increased quantity of health problems impose a still greater challenge for natural absorbers. This has led to the development of artificial ways for shock absorption by either improving or supplementing natural systems. Different approaches have been used for measuring the properties of shock absorbing systems. In this context, shock absorption and attenuation has become one of the most relevant aspects of knowledge in biomechanics. Nevertheless, a better understanding of material's role and heel pad mechanics and of the different testing methods available are needed for further progress-in shock absorption.

3

Comparison of shock absorbing systems properties measured by machine and human testing

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The literature (reported in previous Chapters) contains conflicting results on the shock absorbing properties of materials with respect to both footwear and ~~on~~ the role of natural systems and this makes it difficult to progress the understanding of shock absorption. A great variety of methods for evaluating shock absorption are reported with contradictory results due to unexplained differences in the methodological approach. This thesis addresses these differences and to do this a revision of existing methods was necessary.

Two main approaches may be adopted to evaluate the efficiency of any shock absorbing system: **human testing and machine testing**.

Human tests aim to evaluate the effect of the materials when people wear them in their shoes. In this way, human tests provide information on the effect of shock absorbing materials in the transmission of impacts to the human body. On the other hand, **material tests** use **machines** to measure the mechanical properties of the materials and the results of these tests are then extrapolated to establish the efficiency of the materials in reducing heel strike impacts.

The methods used depended on the nature of the system and on the properties to be analysed. In most cases, the comparison between human and machine methods will provide deeper understanding on the system's performance.

In general terms, human tests are advisable when it is important to analyse the role of the human. Material testing will be used when the effect of the human is either not relevant for the study or already known. Previous machine testing set up procedure normally requires validation by human testing. Material testing may be a quick and easy way of characterising material' performance under in-use-conditions and will inform materials' selection when there is a great variety available. Thus, a reliable and simple method for materials testing is essential in shock absorption research.

For measuring the properties of materials and of passive natural systems both approaches are valid. By contrast, human testing is necessary for analysing active natural systems.

3.1. Machine testing

Machine tests measure the mechanical properties of materials independently of the human. The results of these tests are then extrapolated to establish the efficiency of

the materials in reducing heel strike impacts (Nigg & Bobbert, 1990).
Machine tests have the advantage of being simpler and quicker to use, because they are not subjected to the intrinsic variability of human testing. Many different methods have been used for assessing shock absorbing efficiency of underfoot materials. Machine tests found in the literature can be divided into those that measure some parameters such as force, acceleration or deformation (Calder & Smith, 1985; Clarke & Frederick, 1981; Frederick et al., 1984) and those that study the stress-strain characteristics of the materials (Cook & Kester, 1985; Gross & Bunch, 1989; Misevich & Cavanagh, 1984). Some shortcomings are found in the former. Parameters measured during the impact depend on the testing conditions and method that comparison of results difficult. In most of these tests, only the peak magnitudes are considered which also has the disadvantage of losing information that could be useful for understanding the shock absorbing ability of materials (Nigg & Bobbert, 1990).
The materials studied are viscoelastic and non-linear, so their rigidity and ability to dissipate energy depend on frequency and magnitude of loading. This means, for example, if rigidity is studied with loads of lower frequencies than found in practice, a material could seem soft when it is actually hard under real conditions (Gross & Bunch, 1989).

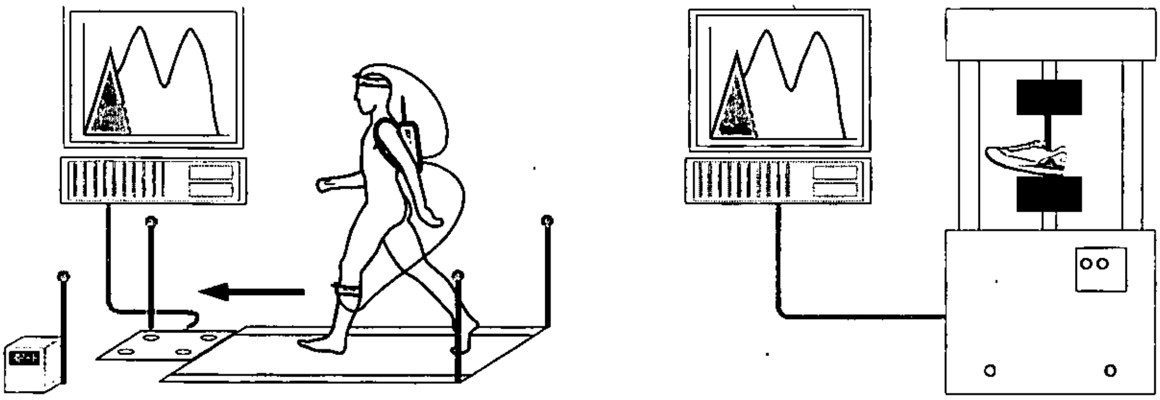

Figure 3.1. Machine testing of footwear simulating real loading

Machine tests, therefore, should apply loads similar in magnitude and frequency to those occurring in real conditions if the non-linear and viscoelastic nature of the materials is to be studied.
The most extended shock absorbing test involves dropping a projectile of a known mass from a given height (Calder & Smith, 1985; Clarke & Frederick, 1981; Frederick et al., 1984). As it is an open loop test, the forces to which the material is subjected are not controlled and depend upon the characteristics of the material. Several authors (Pratt et al., 1986; Shiba et al., 1995) performed drop tests on a material sample

placed over a force plate and the vertical force was registered for assessing shock absorption ability of the material.

Lewis et al. (1991) adapted a resiliometer for drop testing of insert materials. Shock absorption capacity (SAC) was defined as *(maximum deceleration with insert less maximum deceleration without insert)/maximum deceleration without insert*. A performance index (PI) was defined as *maximum deceleration^2 /(time of maximum deceleration x energy return ability)*.

Other methods for material testing include the Shore A hardness determination (Nigg, 1986) or force-deformation characterisation using testing machines at lower loading rates, less than those occurring in real situations (Cook & Kester, 1985; Gross & Bunch, 1989; Misevich & Cavanagh, 1984).

Most recent interest in viscoelastic engineering materials focus on the influence of temperature and humidity as well as long term durability of properties (PFI, 1988; Pratt et al., 1986). Pratt et al. (1986) used drop test and a ballistic pendulum to assess impact properties of some shock attenuating insole materials at in-shoe temperature and humidity. The drop tests were performed in an atmospheric chamber to control the humidity and temperature of a material sample. The pendulum tests were performed with a human sitting with the leg horizontal and his back up against a wall and an accelerometer attached to the tibia. The material under test was placed inside the participant's shoe. More applications of ballistic pendulum for footwear testing are described in Chapter 7.

Kinoshita et al. (1996) measured the effect of environmental temperature on the properties of running shoes. EVA midsoles were tested using a drop test at simulating midsole temperature during running. Different hardness midsoles were tested in a temperature control chamber from -5 °C to 55 °C. Results indicated that shock attenuation of running shoes with EVA midsoles falls with decreasing temperature.

Pirmasens Footwear Institute of Germany (PFI) (1988) pointed out the frequency dependence of shock absorbing properties of shoe bottom materials and developed a stress-strain measuring device pneumatically driven to study the effect of repetitive loading on materials properties. Results showed that the materials' properties degrade under even short wear.

3.2. Human testing

The aim of the human tests is to elicit the material's performance in walking. People influence shock absorption because they interact with the footwear and because of their accommodation to walking conditions. Accommodation depends on the humans' natural responses, all of which should be considered analysing the results of these tests.

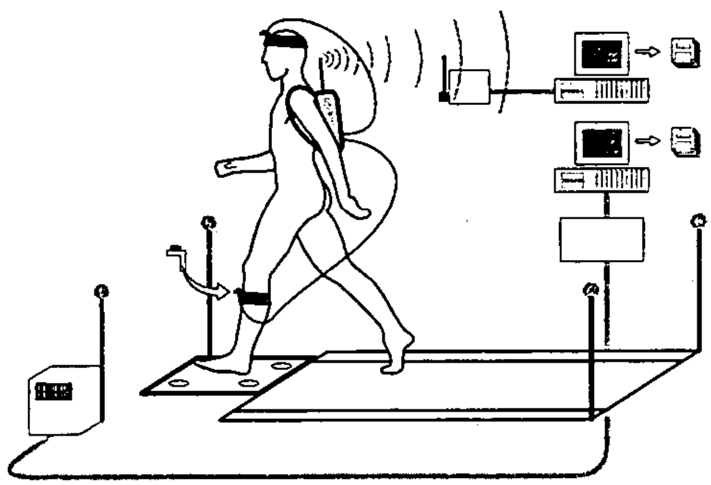

Figure 3.2: Laboratory set up for the recording of impacts during gait.

The human body can be considered a mechanical system. In heel strike an impact force acts as an input to the system, giving rise to a stress wave that is transmitted through the musculoskeletal system up to the head. Therefore, in human testing, two groups of biomechanical variables - system input and output - can be measured: Ground Reaction Forces (GRF) corresponding to heel strike and acceleration experienced in some parts of the body are related to shock wave transmission.

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The two facilities most commonly used to evaluate the *in-vivo* cushioning properties of footwear at heel strike are force plates and accelerometers (Figure 3.2). Force plates measure the external forces while accelerometers measure the characteristics of the shock caused by the external forces (Collins & Whittle, 1989; Lafortune & Hennig, 1992). The combination of both methods is almost universally used to study shock absorption and attenuation of insole materials during walking (Folman et al., 1986; Lafortune & Hennig, 1992; Pratt et al., 1986; Van Jaasveld et al., 1990). Milgrom et al (1996) developed an alternative technique to measure tibia stress by rosette strain gauges attached directly into the tibia of volunteers to analyse footwear influence in stress fractures.

In this way, human tests provide not only information on force reduction but also on the transmission of impacts to the human body. The main disadvantages of these methods are the intrinsic variability of people, the cost and time and their methodological complexity.

A force plate can be used to measure the ground reaction forces (Hennig & Lafortune, 1991; Hennig et al., 1993). Force platforms have been used to define the magnitude and direction of GRF applied by the foot to the ground. A distinctive "M" shape of the vertical force curve is observed (Figure 2.1). This signal is the result of the superposition of two components: the support of the weight of the body and the vertical acceleration of the body. An additional force peak, the heel strike transient, is observed during the initial 50 milliseconds of the signal. In the past, there was controversy over the presence of the heel strike in the vertical component of the GRF. This was due to low pass filtering, too low sampling frequency and low resonant frequencies of the force plate. Nevertheless, with a force platform of sufficiently high natural frequency, a sufficiently large sampling rate and without excessive low-pass filtering, the heel strike transient is visible in the majority of force platform recordings and it is accepted that the heel strike appears as a peak in the vertical forces at the beginning of the contact phase (Figure 2.1) (Light et al., 1980).

Jorgensen and Ekstrand (1988) used force plates to study heel pad confinement by a heel counter and an anatomically formed insole in walking test. Walking test reflected shock absorption increase with heel pad confinement, the effect is greater in low impact loading conditions, probably due to non bottoming out of heel pad. Conflicting results regarding the cushioning properties of footwear during walking have been provided by force plates (Cavanagh et al, 1981) and seems to be related to shock absorption mechanics rather than to force plate technique. Transmission and some characteristics of the transient stress waves generated at heel strike can be measured with accelerometers. Accelerometers have been widely used to measure shock waves travelling through the musculoskeletal system (Johnson, 1986; Light et al., 1980; Nigg & Herzog, 1994; Valiant et al., 1987; Voloshin & Wosk, 1981, 1982; Smeathers, 1989). To accurately measure bone acceleration, it would be necessary to directly attach the accelerometer to the bone as in some reports when an accelerometer was mounted onto a pin inserted into the bone of volunteers (Lafortune & Hennig, 1992; Light et al., 1980).

Since this method presents several disadvantages (Light et al., 1980), many efforts have been attempted to validate the measurement of the bone acceleration by means of a skin-mounted accelerometer. Rooser et al. (1988) used accelerometers attached with an elastic strap onto the skin both at the tibial tuberosity and anteriorly 5 cm. above the medial malleoulus to study deceleration forces in tibia in rheumatoid patients with a knee arthroplasty.

Voloshin et al. (1981, 1982, 1993b), have widely used low mass accelerometers to study shock wave transmission as a method of evaluation of the shock absorbing capacity of viscoelastic materials during walking. In 1994, Voloshin and Loy (1994) used a low mass skin-mounted accelerometer with a velcro strap to evaluate the amplitude of the shock waves caused by wearing high heel shoes during normal gait. They also studied the effect of heel height on the dynamic loading of the human musculoskeletal system and to study the modification of shock waves using a variety of viscoelastic heel plugs and insoles inside the shoe. Impulsive loading was shown to increase with heel height, surpassing barefoot loading when 45-55 mm high. Use of viscoelastic insoles and heel plugs significantly attenuated impulsive loading in high heel gait. In another work (Voloshin et al., 1985) used a force plate together with an accelerometer attached to the tibia with a commercially available knee brace for the analysis of leg transfer function in frequency analysis.

One of the first studies to verify the accuracy of skin-mounted accelerometers to measure actual bone motion was carried out by Saha and Lakes (1977) who evaluated the effects of skin and soft tissue on accelerometer outputs. They reported two main conclusions namely (1) with no preload force applied on the skin-mounted accelerometer to compress the layer of soft tissue, the measure does not accurately correspond to the bone acceleration and (2) the properties of the soft tissue between bone and transducer must be considered when attempting to measure bone vibration.

Ziegert and Lewis (1979) studied the effects of soft tissue by comparing the output of a needle-mounted accelerometer directly attached to the bone with that of a skin-mounted accelerometer. They found that soft tissue effects were negligible under adequate load conditions and low mass. Light et al. (1980) compared the outputs of bone mounted and skin mounted accelerometers and concluded that ... " recordings from the skin surface are broadly similar to those from the tibia, but show loss of high frequency components and some overswing. They were adequate enough to give the order of magnitude of the transient, but not the rate of loading ".

The most recent efforts have attempted to calculate a correction coefficient to the acceleration obtained from a skin-mounted accelerometer. Valiant et al. (1987) obtained a magnification factor that can be used to correct the amplitude of the skin measurements, since its signal shows overswing. Trujillo and Busby (1990) introduced a model similar to that of Ziegert and Lewis (1979) and applied numerical methods to correct the signal; however, they did not carry out experiments to validate the results. Finally, Kim et al. (1993), using a method similar to the above proposed a function to correct the signals based on the measurements of skin-mounted and bone-mounted accelerometers as output and input to the system. However, the main problem is how to evaluate the mechanical parameters of the soft-tissue layer without employing invasive methods.

In all cases, it is important to remember when using accelerometry that the accuracy of the recorded measurement will be dependent upon the attachment of the accelerometer. To summarise literature recommendations, the fundamental requirements to properly measure bone vibration by means of skin-mounted accelerometers are the following:

i. To choose an *appropriate location* so that a thin layer of soft tissue is interposed between the accelerometer and the bone; (Saha and Lakes, 1977)
ii. To use a *lightweight accelerometer*, (Kim, 1993; Ziegert and Lewis, 1979)
iii. To *tightly wrap* the accelerometer to the skin (Kim, 1993; Saha and Lakes, 1977; Ziegert and Lewis, 1979).

Time analysis of force and acceleration has been usually used for heel strike analysis. Frequency analysis of force and acceleration has been also used and some interesting results have been obtained using this technique. Fourier Analysis of force plate and accelerometer recordings showed a high frequency content for heel strike (from 100 Hz to 130 Hz) (Folman et al., 1986). Johnson (1988) developed a parameter called the Shock Factor to evaluate shock attenuation from a device called the JP Biomechanics Shock Meter based in lower limb accelerometry. Stress wave severity is defined in frequency domain by amplitude and frequency content. The efficiency of a given material could be defined as its ability to reduce high frequencies and amplitude. The Shock Factor was defines as $S = As/(An+As)$ where As is the area between the spectrum from 50 to 150 Hz and where An is the area from 10 Hz up to 50Hz. Efficiency of a material will be given by reduction in shock factor (most common footwear materials show about 30% reduction of shock factor). Pratt et al. (1986) used this method to evaluate the effectiveness of insole materials. Effectiveness was defined as Shock Factor reduction with respect to non-insole condition.

According to Voloshin & Wosk (1982) the amplitude variation of signals recorded by accelerometers can arise from several causes other than skeletal damping. In fact, differences between walking conditions have been observed both in acceleration amplitude and shape that suggested the idea of using frequency analysis for the study of recorded signals. Fast Fourier Transform was used for frequency analysis and a transfer function was calculated for the leg from the foot to the knee. The typical transfer function for the lower leg of a healthy young male showed a peak about 20 Hz that would indicate that frequencies outside the range of 20 Hz were more efficiently attenuated. Patients showed after tibia fracture that the main peak frequency shifted to at least 10 Hz higher.

Collins & Whittle (1989) referring to works of Voloshin and Wosk (1981) based their criticism on the fact that attenuation was calculated as the ratio of peak acceleration at different locations and gave little information about ratio of input energy to energy dissipated. However, it is possible that elastic storage, dissipation and dispersion all contributed to the observed reduction in peak magnitudes. More information is needed for a better understanding of the components and strategies possessed and utilised by the human body to withstand and manage impact forces. In most cases, the shape of a stress wave will change as it is transmitted through a body. Dispersion is due to the fact that the speed of a wave through a body varies with its frequency content. Wave components of different frequencies have different speeds through the material. Thus a wave could be dispersed through the material with no energy loss.

3.3. Comparison of results obtained by machine and human testing

Human tests do not always correlate with machine tests (Nigg & Bahlsen, 1987). Differences have been found between properties of materials obtained in machine testing and performance of materials when walking and running. Although some proposals have been considered, these differences have not been fully explained. In spite of extensive research effort devoted to this topic, reliable relationships have not been established between biomechanical measures of impact force developments during running and walking in shoes and the mechanical impact response of the shoes. As a result, even though a great effort has aimed at assessing performance of viscoelastic materials and natural systems there still remains some controversy on their actual shock absorbing effectiveness. Some authors have reported good correlation between drop test results and acceleration recordings obtained from running and walking tests (Pratt et al.; 1986; Wilson, 1985). However, others have showed poor correlation. Shiba et al. (1995) found 11% differences in material properties between drop test (golf ball drop onto a material placed over the force plate) and human testing. Hamill et al. (1996) found lack of correlation between physical test of cushioning and performance tests of materials. Material testing shows that increased hardness increases impact forces and decreases time to peak force. Testing the same materials with humans resulted in different relationship between impact forces and material hardness (Lees, 1988; Robbins and Gouw, 1990; Shiba et al., 1995).

Thus, a simple and reliable method to evaluate and select materials predicting properties in use would be of great utility. However, currently human testing (longer and more expensive) is required to test actual effectiveness of materials. It is widely accepted that providing an explanation for these differences would greatly advance the analysis of shock absorption. In the beginning, some authors attributed that lack of correlation to imprecise recording of the impact event (Shorten & Winslow, 1992) but this has now been overcome and several other explanations are considered.

3.3.1. Research issues

For ten years researchers have proposed different explanations for these differences which in the turn have opened new fields of knowledge but have not yet reached a clear conclusion and this is the future challenge for biomechanics of shock absorption. The main accepted explanations are:

- *No accurate simulation of real loading conditions in machine testing.*
- *Bottoming out of materials in use and degradation of properties*
- *The influence of footwear on walking kinematics and that heel pad confinement has greater effect in shock absorption than materials contribution*
- *Humans' accommodation to walking conditions*

3.3.1.1. No accurate simulation of real loading conditions in machine testing.

According to Lewis et al. (1991) there is an important oversight in establishing a machine test that would comprehensively characterise the properties relevant to the performance of viscoelastic materials. Most of the materials used for inserts and footwear manufacturing are both non-linear and viscoelastic and their properties depend on the rate and magnitude of loading (PFI, 1988; Whittle, 1997; 1999). In consequence, the results obtained in such tests may exhibit a poor correlation with the properties of the materials under heel strike. As an example, this may happen when the impact is much faster than the heel strike - as in the drop test - or when impacting is too slow - as in some Instron tests performed at only a low frequency. Since the rigidity of a viscoelastic material increases with frequency (Gross & Bunch, 1989), an excessively fast impact would yield rigidity measurements above those corresponding to heel strike. In the case of impacts that are too slow, the opposite would occur, i.e., measured rigidity would be lower than expected. Similar considerations apply in the case of impact magnitude. Increasing the magnitude of the impact force characteristically increases the rigidity of a non-linear material. Consequently, if the material is subjected to a lower impact forces the rigidity measured will be less than that obtained with forces similar to heel strike. Depending on the degree of non-linearity and viscoelasticity of the materials studied and if the applied forces fail to reproduce heel strike then not only the rigidity values may change but also the ranking of materials studied with respect to this parameter.

This could be the reason for discrepancies since a limitation of current machine methods is that they do not simulate real walking or running impact loading. Only few methods reported in the literature have subjected the materials to forces similar to those occurring during the movement studied (García et al., 1992; 1994; Swigart et al., 1993) and the results of these methods have not been correlated to human tests.

To explore this statement, a comparison between human testing and a new machine test simulating walking impact loading would be required.

3.3.1.2. Bottoming-out of materials in use

An undesirable property of many foams and "structured" materials is "compression set", in which repeated loading causes structure collapse and the failure to recover after loading release leading to a change of mechanical properties in use (Whittle, 1997; 1999). In this context, bottoming out described the condition when the possible compression is greater than the effective thickness of the material leading to lateral flow and increased material stiffness. This has been regarded as a possible cause of machine and human testing differences (Foti and Hamill, 1993; Hamill, 1996). Foam materials differ in their resistance to compression set due to their chemical make-up and their initial stiffness as does the heel pad and any material with similar structure.

Methods and mechanical parameters are needed to evaluate the degree of bottoming out and its influence in human testing. However, it is very difficult to measure bottoming out of heel pad and materials when walking with footwear. For this reason, indirect methods are sometimes required.

3.3.1.3. The influence of shoe in walking kinematics and heel pad confinement has greater effect in shock absorption than underfoot materials.

Test results that show that force magnitude is independent of shoes and this has been explained in terms of heel pad confinement and kinematic control exerted by the shoes and that these were responsible for changing force magnitudes not the sole materials. Insole materials, however, promote reductions in acceleration magnitude (an additional

10% to add to 70% reduction due to footwear over barefoot). The confinement of the heel pad by means of a rigid counter in the shoe has been showed to be very effective on degenerated and injured heel pads (Jorgensen & Ekstrand, 1988, Jorgensen & Bojsen-Moeller, 1989a) augmenting natural shock absorption by up to 50 %. These results tend to confirm that footwear design plays an important role in shock absorption during locomotion by modifying natural shock absorption but more information is needed on heel pad impact behaviour.

Nonetheless, contradictory results are found in the literature on the effectiveness of heel pad confinement and little knowledge on heel pad impact mechanics and ~~on~~ its dependence on personal characteristics. To advance in this research issue an in-depth analysis of heel pad mechanics is needed.

3.3.1.4. Accommodation to walking conditions

Hamill (1996) suggested that people adjust their kinematics to help attenuate the impact force. The strategy would be individual-dependent and Collins & Whittle (1989) pointed out the need of measuring kinematics when studying impact forces. In the same context, some authors have developed an important theory about human perception of impact as a mechanism for adjusting movement strategy to walking conditions according to a controlled feedback system (Lees, 1988; Robbins et al., 1988a; 1988b; 1993; Robbins & Gouw; 1990; 1991). This is the most complex statement and is currently the most actively investigated one because it has introduced new concepts such as impact perception, passive and active interaction which will be explained in Chapter 4. It requires a more complicated range of ideas to explore it.

This chapter so far has described the differences in results found in literature when shock absorbing systems are tested with people and when tested with machine without people involvement. The conclusion is that the machine measured properties of shock absorbing materials fail to predict their actual performance during walking. Although various explanations have been proposed, there is no general agreement and there are four new research fields currently under investigation. There remain many unknowns and an acceptable solution would advance the development of shock absorption. That humans accommodate to walking conditions depending on impact perception is the most oustanding explanation. Literature on this subject abounds on new concepts and knowledge that need to be explained for a better understanding of the present thesis.

4

Proprioception and perception of impact loading

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The idea of accommodation to impact loading is not new (Hennig et al., 1996) but has recently attracted the interest of biomechanicists as one of the most likely source of the differences between human and material tests. At the moment, impact perception and proprioception are important fields of research encompassing many new concepts and ideas that are revised in this Chapter. This thesis attempts to bridge some of the gaps in the knowledge apparent from the literature survey presented earlier. At the same time, new concepts were introduced as a result of this work.

4.1. Introduction

Accommodation is reflected as a kinematic adaptive response of humans that, it has been suggested, is the result of human proprioception driven by perception of impact (De Koning et al., 1993) to protect the body from high impacts (Hennig et al., 1996) according to a feedback controlled model (Robbins & Gouw, 1991). The mechanism of kinematic adaptation is not clearly understood which may partly explain why the understanding of impact mechanics during locomotion has not progressed as rapidly as might be expected from over 20 years of research by biomechanicists (Lake and Lafortune, 1998). However, this research has led to the development of new concepts such as impact perception. Nevertheless, analysis of these concepts has identified many unknowns with respect to methodology and knowledge.

4.2. Theory of Impact Proprioception

The theory of proprioception as a mechanism to regulate impact loading attenuation was developed from Robbins's works on impact perception. (Robbins et al., 1988a; 1988b; Robbins & Gouw, 1989, 1990, 1991). A review of the evolution of proprioception can be found in the paper of Robbins & Gouw *"Athletic footwear and chronic overloading: A brief Review"* (1990).

Robbins (1988-1991) research was based on the observation of the paradox that modern's athletic footwear had failed to prevent or avoid injuries among runners whereas barefoot populations showed a very low frequency of running related injuries. Athletic footwear has been usually designed to protect that fragile lower extremity from chronic overloading yet the incidence of running-related injuries is increasing despite the growing sophistication of modern shoes (Robbins & Gouw, 1988a; Robbins & Gouw, 1990, 1991). It could be that modern shoes for some unknown reason give rise to greater impacts (Robbins and Gouw, 1990) or, perhaps, there is something in barefoot populations activities that reduces the incidence of running-related injuries, or maybe a combination of both.

The first experiments by Robbins on barefoot activity (Robbins et al., 1988), suggested that plantar sensory feedback may induce intrinsic foot shock absorption by means of an avoidance behaviour caused by noxious plantar skin sensation due to irregular natural surfaces. The pain pressure threshold was different depending in the foot plant areas. Thus, under every foot plant locations there would be distinct sensations produced by pressure causing intrinsic foot shock absorption to alleviate pain in the less tolerant areas (Robbins et al., 1988a). This was supported by the results from an experiment (Robbins & Gouw, 1989) using a penetrometer with a spherical end of different diameters to manually press at the three plantar locations of 100 volunteers and of ten fresh cadavers (Figure 4.1). Load varied from 1.8 kg to 9 kg and was applied for 1 second duration. The indentation depth was measured whilst the 100 participants were asked whether they experiencing pain.

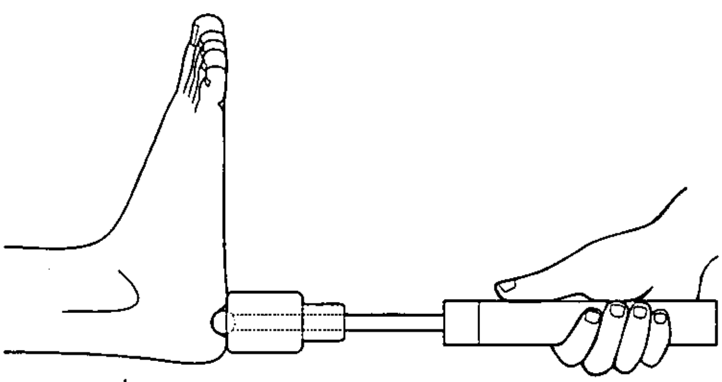

Figure 4.1. Experimentel penetrometer set up used by Robbins

Results showed that sensory thresholds for pain were higher at the heel pad than in the distal first digit and lower in the metatarsal phalangeal joint. These results have been confirmed by several research projects conducted by the writer and others at the IBV using a similar methodology with adults and elderly people (Gonzalez et al., 1999). In this method a commercial dynamometer with an aluminium end was used to manually press at different plantar locations until participants stopped the test due to discomfort (Figure 4.2).

So, plantar load redistribution to release the forefoot would be achieved by muscle activity including longitudinal foot arch deflection and hip flexion (Robbins et al., 1988a, 1989) both of which control temporal aspects of impact loading (Robbins et al., 1989).

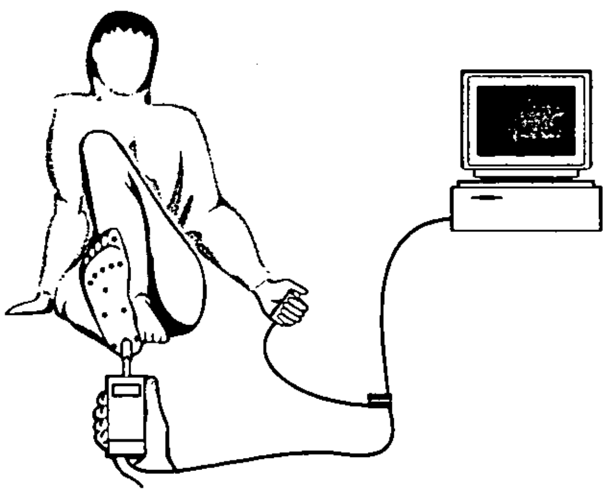

Figure 4.2. IBV method to measure pain threshold at the foot plant

Robbins *et al.* (1988a) stated the hypothesis of the existence of barefoot adaptations based on a load magnitude sensory feedback control system that prevents overloading during locomotion providing impact absorption and protection against running-related injuries. This human natural capability would be masked by modern athletic footwear. This is supported by experiments on load estimation (Robbins et al., 1988a) combined, possibly, with mechanical interference in arch deflection due to shoe features such as lacing and arch supports resulting in running related injuries (Robbins et al., 1988a).

Later works by Robbins et al. (1988a, b) emphasised the importance of surface material and roughness in relation to plantar pressure detection for load estimation to start neuromuscular adaptations necessary to avoid chronic overloads. In these experiments (1988a) significant differences were found in flat and rigid surfaces varying only in texture. The results showed that the magnitude of the protective avoidance behaviour varied as a function of the underfoot-contact surface. This implied that there was a control system that could modulate the magnitude of a load experienced on the plantar surface and it was influenced by factors affecting perceptual processes. In particular, these findings suggested that the perceived magnitude of an applied load might be an important factor in determining when various protective mechanisms become evident during running. In these experiments, participants were asked to quantify the estimated perceived load on the plantar surface from variable loads ranging from 0 kg to 164 kg applied in participant's knee while seating knees bent at right angles and the foot resting on different surfaces (gravel, smooth plastic and shoes) (Figure 4.3).

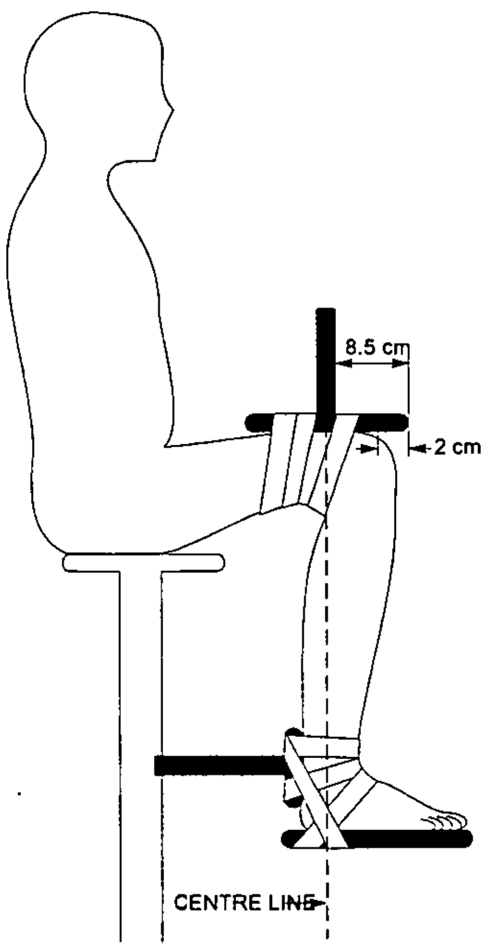

Figure 4.3. Robbins method to analyse load estimation

These results reinforced the importance of sensory-induced avoiding behaviour and plantar surface sensitivity for pressure detection associated with the physical interaction of the plantar surface with the ground (in the unshod), or the footwear and underlying surface (in the shod) as well as the influence of footwear on it. Thus, a lack of plantar skin sensation in excessively cushioned running footwear could create a perceptual underestimation of actual impact severity thus provoking injuries (Robbins & Gouw, 1989, 1990, 1991). Another conclusion s the importance of registering plantar pressure distribution in walking tests of impact proprioception.

Robbins et al (1991) evolved the theory that in humans the avoidance of uncomfortable or painful, but locally innocuous, plantar cutaneous tactile stimuli moderates shock on subsequent impacts when walking, running, or jumping repetitively. This feedback control circuit is optimised in terms of protection for mechanical interaction of the barefoot and natural surfaces allowing for anticipatory avoidance. Modern athletic footwear would turn unsafe because it attenuates plantar sensation that induces the behaviour required to prevent injury.

This theory could explain the difference in injury incidence between barefoot and shod runners via the requirement of plantar discomfort on impact for optimised shock absorption. It could also explain why material tests fail to predict actual impact when running. A more compliant shoe, which according to material tests should attenuate shock more effectively, fails to do so because it produces greater plantar comfort, hence less impact-moderating behaviour. Results on relationship between surface irregularities and load perception as well as arch rise and hip flexion support this theory (Robbins et al., 1988a, 1988b). In conclusion, when plantar sensory consequences of impact are attenuated, humans underestimate impact and reduce impact-moderating

behaviour, which may elevate impact sufficiently to cause chronic overloading (Robbins & Gouw, 1991). In the sole of the foot the relationship between the magnitudes of applied and perceived force depends on the material in contact with the foot (Robbins et al., 1988). Thus, perceived force is greater for materials with irregular surfaces than for those with smooth surfaces. As a result, the participant underestimates the impact with certain surfaces. This phenomenon has been called "Discomfort Impact Illusion" (Robbins & Gouw, 1991) and a number of studies suggest that it affects both the perception of impact and adaptation to it (Robbins et al., 1993).
The masking effect of modern athletic footwear, described above, produces so called *"discomfort impact illusion"*. Similar results have been obtained for landing on hard and soft surfaces (McNitt et al, 1991).
The theories and experimental results of Robbins & Gouw (1991) have been summarised in the *proprioceptive model* on which a great part of shock absorption research is currently focused (Figure 4.4). In this model the human body is described as a feedback control system. The human body adapts to the circumstances of locomotion (Clarke et al., 1983a, 1983b) determined by displacement velocity, the type of ground, footwear etc. The interaction between the humans and the circumstances of locomotion may be described by means of a proprioceptive model and how these circumstances create an input that is recognised by the feedback control system. The control system responds to the signal and generates a response to correcting possible discrepancies.
In such a model three different functions can be identified:

i. A control system which compares input with its goals
ii. Response (output) generated to modify the input according to system objectives
iii. Sensing of Input (locomotion circumstances) which is transmitted to the control system (perception).

The system is obviously made up of biological and physiological materials and structures. The mechanical sensory systems involved in the detection of shock related quantities are diverse (skin receptors, joint receptors, vestibular apparatus, muscle spindles). However, impact perception seems to be more related to sensory processes of the skin than to muscle receptors. In this context, Robbins and Gouw (1991) suggested that the polymodal nociceptors with C-fibre afferents are well suited for impact sensing. Given that their threshold depends on displacement and superficial deformation of the skin, they show an easy sensitisation and their temporary response appear to be well adapted to the dynamics of locomotion. However, this aspect of the system is outside of the scope of this thesis. The Robbins's papers (1990–1991) have a comprehensive physiological description.

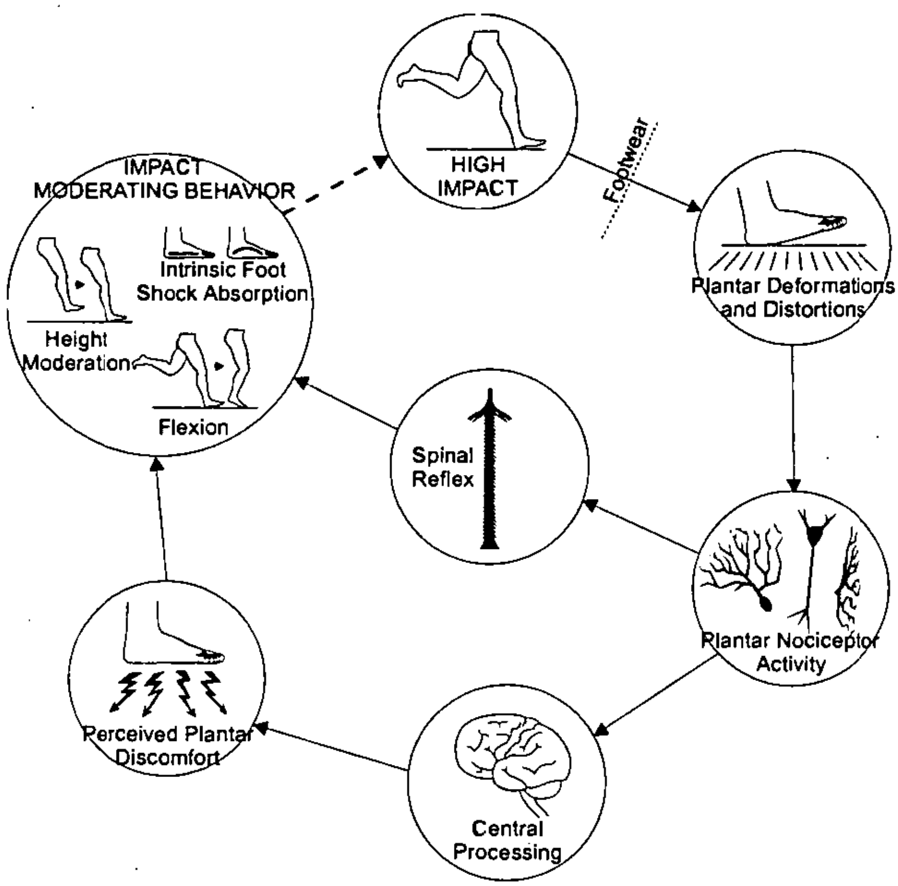

Figure 4.4. Robbins feedback controlled model.

This new model caused an important change of direction in shock absorption research. Each of the model functions present new challenges and has identified important gaps in current knowledge and understanding and will be the focus of future research. Already, there is much recent research activity devoted to impact perception.

4.3. System Output and Accommodation Mechanisms

The response of the proprioceptive system consists of a variety of mechanisms that humans use for modifying impact loading to avoid the undesirable effects of impact forces. These include both active kinematic and neuromuscular adjustments of the locomotor system to the expected severity of the impact.

This active adaptation has often been suggested to be the mechanism of impact force regulation in running, but there is no strong evidence supporting this (Clarke et al, 1983a; Cole et al., 1995; De Koning et al., 1993; Lees, 1988; Valiant, 1995). Furthermore, there is little known research on impact accommodation in walking (Forner et al., 1995).

These systems include knee flexion, contralateral flexion of the hip, and control of the position of the foot at the moment of heel strike (McMahon et al., 1987; Robbins et al., 1988). Clarke et al. (1983b) described an increase in maximum knee flexion velocity as an adaptive response to an increase in the relative hardness of the shoe. This type of response is identified as kinematic adaptation and was later described by Lees (1988) as the result of stereotype movement generated by humans under a specific set of environmental conditions. Adaptive lower limb kinematic adjustment has been demonstrated when changing from shod to barefoot locomotion (Koenig, 1993) and during locomotion in footwear of different midsole densities (De Koning et al., 1993; Lake & Lafortune, 1998).

Nevertheless, in the literature there is some discussion about the relative contribution of mechanical and neuromuscular factors for controlling the impact conditions. In recent work using human pendulum testing, Lake and Lafortune (1997) examined the muscular preparation for lower limb impact following an unexpected change in the midsole density of athletic footwear. Results on wall forces, shank acceleration and electromiography (EMG) of lower leg muscles showed that the severity of lower extremity impact was modified mainly due to the mechanical characteristics of the shoe midsole whilst the effects of short term muscular adaptations were minimal or absent when changing from hard to soft shoes and vice-versa. Precontact muscular activity of selected lower extremity muscle was not modified after the change in footwear or after accommodation by treadmill running. The response to a switch in footwear appeared to be mainly mechanical in nature and dependent upon the degree of change in midsole properties of the test shoes. The absence of change in muscular preparation prior to impact was explained by the authors as due to that the unexpected adjustment in impact severity was either not perceived by the participants or they deemed such a change as unnecessary. Muscular preparation for impact loads, over the range experienced in that study, appeared to be relatively independent of shoe midsole mechanical properties. Although the participants were allowed to use the shoes before testing for accommodation, the testing procedure was static and thus restricted the possibility of muscular response due to kinematic accommodation to changes in impact conditions which in walking and running defines muscular preparation for heel contact. In a passive test muscular preparation is probably more limited and a fine adaptation not possible. In fact, some works revealed midsole cushioning differences only when kinematic adaptations were restricted and the body was forced to respond in an other mechanical manner (Lawless & Lafortune, 1997). In this sense, some authors have demonstrated adaptive strategies to specific mat conditions for drop landings in such a way that associated changes in impact severity alongside the neuromuscular adjustments were found (Anderst et al., 1993; Lawless and Lafortune, 1997; Sanders and Allen, 1993). Studies on response upon unanticipated perturbations of the conditions of impact (Dyhre-Poulsen, 1984; McKinley; 1983) have established that the onset of muscle activity is planned in advance and based upon the severity of prior impacts.

4.4. Input sensing. Perception of impacts and footwear cushioning

Early work by Robbins suggested that the perceived severity of foot-ground impact is likely to be an important factor in the adaptive response of the body to protect against chronic overloading and high impact shocks during locomotion (Robbins & Gouw, 1991). Until the body responds to reduce these loads, the body relies to a large extent on its ability to recognise or perceive their existence (Bates et al., 1988; Caster, 1995; Robbins et al., 1988a). The proprioceptive model is based on the perception of mechanical inputs related to impact and its transmission to the control system for comparison with system's goal. Thus, when the cushioning capabilities of the body and/or foot-ground interface cannot sufficiently attenuate the impact, the role of the sensory faculty of the human body to perceive the severity of impact loads becomes very important in setting up injury avoidance strategies.

Perception of stimulus has been widely studied and phychophysics studies on this particular subject are found in scientific literature of many different disciplines. However, perception of impact loading has only recently received some attention from biomechanicists but the current understanding is far from complete.

It is commonly accepted that some degree of shoe cushioning is important for the protection of the human body by effectively reducing the impact forces and the transmitted shock wave and this has been demonstrated in running, walking and human pendulum investigations (Lawless and Lafortune, 1997). However, according to recent results (Hennig et al., 1996; Robbins, 1988-1991) the perception of cushioning is also an important issue in the discussion of the causation of sport injuries. Thus, not only cushioning but perception of it during human locomotion may play a crucial role in protecting the body from chronic overloads.

Many authors have studied the perception of cushioning, but they have always been interested in running conditions. Limitations in sensitivity have been postulated restricting the perception ability in such a way that impact conditions that are lower than those arising from moderately severe running would not be perceived (Lafortune, 1997). So, perception studies during walking became necessary to contrast the sensibility of the impact perception system.

In general, perception studies in the literature provide little information about the biomechanics of perception, but many methodological questions about rating scales, experimental procedure, testing methods and others are raised. Consequently, the experimental study of this subject comes out to be quite obtuse and complicated and a revision of methodology reported in the literature was deemed necessary. Many questions are raised but the reliability and quality of the answer depended on the methods used to process data. The relationship between perceptual and stimulus intensity can be drawn from judgements of the each participant in various procedures (Stevens, 1957). A number of psychophysical procedures have been used for measuring sensory attributes.

With respect to the input, two main questions emerge from perception studies in the literature (Hennig, 1995a), namely:

i. It is not clear which biomechanical variables govern impact perception and how these relate with systems' objective
ii. It is not known which footwear properties relate to impact perception or how they relate.

Understanding the physical parameters that the human body uses to sense impact perception is very important because these are the variables that biomechanists would wish to measure and may wish to modify to control accommodation and thus shock absorption.

4.4.1. Methodological aspects of impact perception experiments

The major recommendation that emerges from the literature search is that experimental protocol must be as simple as possible (Milani et at, 1995). Some research in the literature showed a low individual correlation between biomechanical variables and perception suggesting that complex demands of the experimental protocol led individuals to rely upon other stimuli to assess impact severity (Milani, 1995). In the work of Milani et al. (1995) for instance, participants were required to perceive impact severity, heel pressure, and rearfoot movement when attempting to hit a force platform at a given velocity. This, of course, demanded a high level of concentration and co-ordination.

The main methodological questions usually discussed in literature are:

i. Rating scales
ii. Use of reference conditions
iii. Provide the participants with the range of impact conditions

iv. Experimental procedure

v. Experimental conditions: passive versus active.

Different *rating approaches* have been proposed in psychophysical studies as useful in identifying relationships between various subjective and stimuli magnitudes (Stevens, 1956, 1957). Two main approaches have been used to study subjective perception of product's aspects: *Likert and category estimation.*

In early experiments perception was measured using a 1 [very poor] to 7 [very good] point **Likert scale** and this scale is still widely used in comfort studies (Corlett & Bishop, 1976; Shackel et al., 1969). In subsequent studies the modified 15 points Borg scale was used.

On the basis of psychophysics, the rating of perceived exertion proved a useful tool in exercise physiology, effort and workload analysis (Genaidy et al., 1998). For sports stimulus rating is used. A critical review on rating scales for perceived exertion was provided by Carton and Rhodes (1985).

Stevens (1956, 1957) pioneered work in the ratio scaling of psychological magnitudes and originated **the magnitude estimation procedure**. Stevens (1956) suggested that, if properly used, this method could provide a simple, direct means of determining a scale of subjective magnitude. This has been the most common procedure employed in psychophysics over the last 40 years (Stevens 1956-1957, 1972a, 1972b).

Borg (1962) used a 21 point rating scale for analysing the perception of effort during aerobic work. In 1970, to increase linearity between rating and workload, Borg developed a 15-point graded category scale for workload studies, ranging in values from 6 to 20. In cushioning analysis, scores 6 to 10 represent increasingly cushioning from satisfactory to very good. Scores 11 to 15 represent progressively unfavourable stimulus due to excessive stiffness and 5 down to 1 represent progressively worse cushioning due to increasing hardness (Valiant, 1995). This method is called the **15-point scale Category Estimation Method** and has become a standard for rating perceived exertion.

The major difference between methods is that whilst Borg and Likert used fixed scales that could influence the freedom of assessment, the "Magnitude Estimation Method" introduced an open scale and participants allocate a number to the stimulus perceived.

In some studies, participants were completely free to assign any positive numerical value to a particular perceived severity of the impact (Lake and Lafortune, 1998) or just on a scale of 20 lb (9.9 kg) increments (Robbins et al., 1988a). Thus each impact had a magnitude assigned to it by the participant. The assumption in this method is that numerical judgements are directly proportional to sensory magnitude. Work on the additivity of measurements seems to indicate that, at least for data averaged over several participants, the assumption is correct (Bolanowsky et al., 1991).

The experimental procedure has also a great influence in the results. According to a definition by Atkinson and Atkinson (1980) perception requires the interaction between sensory mechanisms and those parts of the brain that are concerned with storage and retrieval of past experiences. Thus, perception is a dynamic process influenced by past and present experience that depends not only on the current stimulus but on the expected stimuli (Bolanowski et al., 1991). So, past experience has an influence in perception but fatigue, fear, and other emotions would probably also influence subjective ratings of loads.

Thus, the testing protocol and procedure are very important for the quality of the final results. If testing conditions are unusual for the participant then past experience component may be different. On the other hand, measurements of psychological magnitude have sources of potential bias that must be minimised in the experimental design. Subjective ratings can be biased by conditions within the experiment and specially by characteristics of rating procedures such as participant's instruction, or maybe a product of the participant's past history (Lake and Lafortune, 1998). Psychophysical procedures have been developed so that individual can assess their impressions and communicate them to the experimenter with as few biasing cues, suggestions, and constraints as possible. In this sense, the way of comparing the results between participants are major methodological aspects (Hennig, 1995a). Some of the recommendations made by Stevens (1956, 1957, 1959) deemed important for reducing bias have been successfully incorporated in different studies (Lake and Lafortune, 1998; Milani et al, 1997).

Regarding *the testing protocol,* it is preferable if all impact testing is performed in silence to allow the participants to concentration on perceived severity. The same instructions should be given to all and they must be informed that impact conditions would vary randomly. Participants are normally asked to focus on the perceived severity of each impact. Robbins & Gouw (1991) informed the participants of the experimental objective and the instant when they should concentrate and estimate the load.

So when is rating performed? Some authors have done this during running in shoes and others after a run, or under impact conditions relying on the accumulation of experience for the whole duration of the activity.

It is usual to use a specific standard or "anchor" impact condition that the participants' use as a *reference point* in assigning numerical values to subsequent test conditions.

For Lake and Lafortune (1998) the standard impact condition was assigned a numerical value [10] by the experimenter, chosen because this was easily multiplied and divided. Several approaches have been used in the procedure and analysis of the reference condition. A reference shoe can either be presented only at the beginning of the experiment or it can be tested each time before running with a new shoe model (Milani et al, 1997). This, however, causes problems when testing a large number of different shoes, since if the reference is presented only at the beginning participants progressively lose their reference (past experience component). If it is presented with every new model, testing can get very long and tiring and participants may lose concentration (fatigue influences perception). A compromise is needed. The same reference shoe can be used for each participant but this requires careful selection for the reference condition. This selection enables participants to more easily take "possession" of the standard (Stevens, 1956), for instance, the middle of the range of conditions so that it would not impress the participant as being either extremely soft or hard (Lake and Lafortune, 1998). Similarly for medium velocity and the properties of the materials tested. Alternatively, a reference shoe can be chosen individually for each participant by random selection (Milani et al, 1997). Robbins & Gouw (1991) chose the reference for scaling stimulus by asking participants to select the most uncomfortable surface. The maximum impact on the most uncomfortable surface was assigned a discomfort rating of 100. The most uncomfortable was then used as reference.

It is also common to **provide** participants with the *range of stimuli* by wearing the softest and the hardest shoe (Milani et al, 1997) or testing the extremes of the load to be applied (Robbins & Gouw, 1991). Robbins et al. (1988a) instructed the participants about the range of load to be applied. In this case, it has to be supposed that the criteria (hardness, impact loading) followed the chosen conditions representing the extremes of the range of products as this is significant for introducing differences in impact perception. In this sense, it is better if the range of impact conditions is not large and presented in a mixed order (Lafotune & Lake, 1998).

On the other hand, two main *testing methods* are apparent, using *active* (Hennig et al., 1996; Milani et al., 1995, 1997a, 1997b) and *passive* testing devices (Lafortune et al., 1995b; Lawless & Lafortune, 1997; Robbins & Gouw, 1991). The former consist of registering biomechanical variables when running, walking or jumping whereas the latter use testing devices such as a pendulum to deliver impacts to the participants. Passive methods exclude interference of locomotor ability that could improve perception (Hennig, 1995a) and allow control of impact conditions that would simplify the analysis of perception (Lafortune et al., 1995; Lawless & Lafortune, 1997). However, active testing is more realistic providing the participants with past real information about each testing condition and the accommodation component is allowed to occur before collecting perception and biomechanical variables. Even though some differences are found in the results from both methods, there is no comparative analysis. The research works found in literature basically differ in the methodology used for acquiring biomechanical variables.

Active testing. Milani et al (1995) conducted perception studies during running and unexpected falls comparing ground reaction forces parameters (GRF), peak tibial acceleration, rearfoot movement and plantar foot pressure variables with perception scores using the 15 point modified Borg scale with reference stimuli. Runners' rating were best for the unexpected fall condition on 9 different materials. In this condition they did not have to divide their attention between the running performance and their perceptual task.

Hennig et al (1996) collected perception scores using a modified Borg 15-point scale to judge the value of perception in detecting force-related variables whilst running on a treadmill and compared these with pressure distribution and ground reaction forces registered running over ground afterwards. Peak force, maximum force during midstance, maximum force rate and median power frequency were determined from vertical GRF.

In a similar study, Milani et al (1997a) compared perception ratings of shock, pressure and rearfoot movement with biomechanical variables measured during running and nine shoes differing only in midsole stiffness. Reference and shoe stiffness range was provided. Ground reaction forces, tibia acceleration, rearfoot motion using electrogoniometer and heel pressure distribution using single sensors were recorded as biomechanical variables. In a further study the same authors (Milani et al, 1997b), studied whether small changes in midsole hardness were perceived by participants. Using the same methodology eight shoes that differed only in midsole stiffness under the heel and midfoot area were tested running across a force platform in each shoe at a controlled running speed.

Passive testing. Lake and Lafortune (1998) considered how perception of impact severity could be affected by posture of the foot and lower leg at impact. They concluded that it becomes difficult to determine whether the reported perceptual modifications were the result of changes in either lower limb posture, mechanical inputs or a combination of both when the foot contacted the ground. Isolation and manipulation of mechanical inputs to the body are difficult because of the confounding kinematic adaptation. They concluded that to identify the mechanical basis for the perception of impact severity an altemative controlled *in-vitro* approach was needed to systematically manipulate mechanical inputs to the body. A human pendulum approach was developed to this end since it presents easy control and manipulation of initial impact conditions, along with the possibility of imparting loads that are similar to those produced during locomotion (Lafortune et al, 1995). This technique is similar to the instrumented pendulum used for heel pad testing (Valiant, 1984) but with the subject lying supine on a bed as the missile impacts a wall-mounted force platform (see Chapter 7 for-more details). However, this method did not allow free movement thus limiting past experience was less realistic for a dynamic process as perception. Extrapolation to actual locomotion is difficult because the human pendulum does not reproduce all initial locomotor heel strike conditions. Lake and Lafortune (1998) used this technique in combination with Borg scale with ratio scaling of stimulus and standard reference condition to explore the relationship of the commonly measured biomechanical variables describing impact loading to the perception of impact severity. A force plate mounted on the wall was used to register reaction forces. Accelerometers in the shank and on a bite bar that was firmly gripped between the teeth were used to register acceleration. Pressure on the plantar surface of the heel was also measured using eight discrete sensors. EMG activities of five lower extremity muscles were also recorded.

The wall-mounted force plate was either bare or covered by EVA (ethyl vinyl acetate) foam layers. Nine different test impact conditions, representing all combinations of 3 impact velocities and 3 interface materials Covering the wall-mounted force platform were analysed. The knee was 20 degrees flexed simulating typical knee angle at foot contact during running whereas participants were instructed to maintain a slightly dorsiflexed (less than 5 degree) ankle that ensured a heel first contact, but this required concentration by the participant. Standard condition and trials on the extremes of the range of impact conditions were used to reduce bias. Testing started with 3 impacts at the constant standard condition.

Biomechanical variables measured were scaled for each participant to the mean value recorded for the standard condition. Perceptual rating values were divided by ten so that they were also scaled to the standard condition in the same manner. The relationship between biomechanical input variables and perceptual impact severity was examined using multiple correlation analyses on three different levels by considering all the data or averaged by group or participant.

Significant interaction between impact velocity and force plate interface material was observed for perception ratings. In general, the perceived impact severity significantly increased alongside increases in impact velocity and interface hardness. Lowest impact velocity and less dense interface material revealed no significantly differences between impact conditions. At the same time, the differences in perceived impact severity between the 3 impact velocities were more evident as the interface material density increased. The interface effects on impact perception tended to be larger at the fastest impact velocity (1.2 ms^{-1}). Generally, the perception ratings were similar for the medium and hard interface at a given impact velocity.

The high correlation demonstrated between mechanical input and perceived severity is only in partial agreement with the results of Robbins et al. (1988a). These authors found a strong linear relationship between perceived magnitude and load applied to the lower leg on different surfaces both barefoot and wearing a selection of athletic shoes. However, after extrapolation of their results to typical loads that are experienced during running, they postulated that athletic footwear substantially attenuated the perception of loads so that perceived severity is less than actual. In contrast, this study found a close relationship between perceived and actual severity at a range of impact conditions that were very similar to those experienced during running. This contrasting finding may be partly explained by the fact that Robbins did not simulate time history whereas the human pendulum loads closely simulated the time history during the initial impact phase of running.

In addition to providing a closer simulation of the magnitude and temporal characteristics of locomotor-like loading, the work of Milani et al. (1995) mimicked the angle of the knee at ground contact during locomotion (20°) and isolated the mechanical input stimuli to those generated only at the foot.

The results of human pendulum (Lake and Lafortune, 1998) suggest some confusing statistical interpretation. If all variables were correlated with each other and a Principal component analysis failed to clarify the underlying data structure, it is not possible to say which is the actual relationship because internal correlations may lead to erroneous conclusions. At the same time, participants were instructed to maintain ankle dorsiflexion what demanded high concentration from them and could had bias perception.

Lake and Lafortune (1998) analysed each velocity and minimising the level of expectation of the impact found-that the transient rather than peak variables were more related to the perception of impact severity, as with unexpected falls (Milani et al., 1995). The perceived impact severity was closely associated with the mechanical input variables commonly measured. These findings indicated that midsole materials such as those typically found in athletic footwear (EVA layers) did not remove human ability to perceive the severity of impact loads.

From the literature review, it is not clear whether active or passive testing yield better results. **Important recommendations** for optimising the experimental procedure of perception experiments can be extracted though:

i. A randomly chosen reference condition should be used and presented regularly [every few testing conditions], but without increasing the duration of the testing session.
ii. Test conditions should be presented to the participants in a random or mixed order.
iii. Some test conditions should be presented to the participants before testing to provide an idea of the range of conditions
iv. The testing protocol should be as simple as possible allowing concentration on a single task with a simplified rating scale.
v. A silent and comfortable testing environment should be provided.
vi. Very explicit and similar instructions as to what is required from the participant during testing and the objective of the experiment should be issued to all participants.
vii. Participant's fatigue should be avoided and past experience component of perception allowed.
viii. Any audio and visual cues should be avoided.
ix. Real conditions should be simulated to avoid missing perception factors (dynamic, etc.)

4.4.2. Which input variables control impact perception?

In 1988 Lees (1988) said that research had been unable to relate reliably perception of cushioning of shoes when running to peak impact force. Since then, much research has been devoted to investigate which parameters - or which combination of parameters – can be used to perceive shoe cushioning. Variables as accelerations, forces and impulses on body segments, plantar foot pressures, vibration and even auditory and visual cues may be involved in the subjective evaluation of cushioning properties of footwear (Hennig, 1995a). Some works have shown different relationships of plantar pressures, impact forces and frequency content of loading with impact perception. According to Valiant (1995), both physical measures of cushioning and magnitude of vertical forces typically do not compare well with subjective or to biomechanical measures of cushioning.

Literature search (Lake et al, 1998; Milani et al., 1995) showed that the relationship between biomechanical variables and impact perception generally increased with severity of impacts. Perceived severity became more dependent upon specific variables at the fastest velocity condition. The variables frequently related to perception of impact were characteristic of each person and showed that peak impact force was either moderately or highly correlated with perception of impact severity. In every participant the transient rates of force, shank and head acceleration were generally the better predictors of perception. All of the participants showed at least a moderate association with rate of loading of force, which explained 64% of the variability and head rate of loading with their perception scores. All variables were also correlated significantly.

Pressure distribution was related to load estimation in Robbins's model (1988a, 1991). Plantar pressures surpassing participants' pain threshold were supposed to provoke accommodative behaviour. Different, even conflicting relations between peak plantar pressures and perception of cushioning have been documented. Changes in temporal load distribution under the foot have been observed during running in response to footwear that had modifications in midsole hardness and in which cushioning was perceived very differently (Hennig et al., 1996). The pressure beneath the heel was also related to impact perception. In some studies, the mean peak plantar pressures developed beneath the external heel were progressively lower for shoes whose perception of cushioning became progressively better, similarly, peak pressures under the heel increased in the harder shoes (Hennig et al., 1996; Lake et al, 1994; Valiant, 1995) whilst in others medial heel pressures were more closely related to perception than pressures beneath the lateral heel. Certain peak forefoot pressures were also related to perception: with the perception of reduced shoe cushioning, the participants demonstrated higher forefoot loading (Hennig, 1996). Pressure time integral has been also observed to play an important role in comfort and human perception of shoes (Jordan & Barlett, 1994).

In a study by Hennig (1995a), peak pressure analysis revealed statistically significant differences between shoes under all anatomical locations except for the medial midfoot, the fifth metatarsal head and the hallux. Higher heel and forefoot pressures in hard shoes were contrasted by elevated lateral midfoot pressures in soft shoes. The conclusions of this work should be analysed with care since only three different conditions were tested and more extensive studies would be needed to corroborate these results.

Consistent relationship found by several authors (Hennig et al., 1996; Lake et al, 1994, 1998; Milani et al, 1995; Valiant, 1995) between perception of cushioning and both rate of loading and spectral characteristics of Vertical Ground Reaction Forces could indicate that biomechanical variables related to ***loading frequency*** may be useful objective measures to predict perception.

Nevertheless, whether this relationship validates the reliability of the subjective cushioning measure or, alternatively, identifies an appropriate objective measure has not been confirmed. This point is very important, but the discussion about the relationship between health and cushioning perception is not the goal of this thesis.

Running tests with 4 shoes yielded statistically significant differences in perception which were similar to the statistically significant differences found in average loading rates (Lake et al, 1994). In the study of Milani *et al* (1997) among the different biomechanical variables measured, impact force rate of loading and impact force median power frequency demonstrated moderate to high correlation with the perception ratings. At the same time, regression analysis reflected a high correlation of the median power of frequency of the vertical GRF with impact perception, low of the peak tibia acceleration and high negative of the peak force (R= - 0.73).

Hennig (1995a) tested 3 experimental footwear conditions during running showing that force rate of loading was one of the variables that best correlated to perception of impact severity or cushioning. According to Lafortune et al (1995b) the 3 experimental footwear conditions used had extreme differences in midsole density. By recording in-shoe plantar pressures, these authors found that the different footwear induced the participants to modify the under feet loading distribution and their landing kinematics during running. This adaptation of initial conditions may have confounded the relationship between footwear cushioning properties and perceived impact severity because the adjustments may serve to move perceived 'hardness' to a different or optimum level.

With higher scores of cushioning perception, decreases in maximum force rate as well as the median power frequency were found across the footwear conditions, the latter variable has been related by different authors to shank acceleration (Hennig & Lafortune, 1991; Hennig et al., 1993). Results showed that the transient rate variables demonstrated the greatest change over the range of conditions (>200%) (Lake and Lafortune, 1998).

This relationship has been explained by Milani et al. (1997) as that the median power frequency (MPF) incorporates the amplitude of the different frequency components of the complex force-time signal. Larger MPF values result from increased power components at higher frequencies, therefore a sharper impact will produce a higher

MPF. This suggested that the body's sensory system seemed to differentiate well between impacts of different frequency content. This conclusion, however, would be only valid in case impact sensation was judged by humans through shock transmission sensations rather than by plantar pressure judgement. Nevertheless, the transmission of mechanical transients had already been suggested to be responsible for perception based in the fact that they are adequate stimuli to plantar mechanoreceptors and nociceptors (Robbins et al., 1988b) although some workers found little correlation of shank acceleration with impact perception (Hennig et al., 1996; Milani, et al., 1995).

Other workers suggested that ***biomechanical variables combine*** in a mechanical way to define impact perception. High Correlations have been found between impact forces, pressure and rearfoot motion. This may be due to the fact that softer midsole should be conductive to lower initial impact force, lower loading rate, lower heel pressures at the same time than higher pronation and pronation velocity as the foot accepts the weight of the body. High correlation values were reported between biomechanical and perception variables for shoes with large differences in midsole hardness (Hennig, 1995).

In a similar study, using the same methodology Milani et al (1997) compared perception ratings of shock, pressure and rearfoot movement with biomechanical variables measured during running with 9 shoes differing only in midsole stiffness. Group analysis of correlation revealed that mean force loading rate, median power frequency of force, peak pressure in the heel region, rearfoot angle and maximal angular velocity were highly related to each perception and to each other.

High correlations between the perceptual ratings could indicate that humans used the same or combinations of the same cues for the perceptual ratings, independently of the perception variable. It may also mean that humans were not able to differentiate between impact, pressure and rearfoot instability perceptions. The design of the study made high demands on the concentration of participant and this could have reduced perceptual attention to one single perception variable. For the same reason extremely high correlation between mechanical variables and perceived impact severity using a human pendulum were reported (Lake et al., 1998). But, as reported by Lake et al these may also have been triggered by the much larger range of impact severity, when compared to the relatively small stiffness in the study. Multiple correlation analysis revealed that all the mechanical variables were significantly correlated with perception of impact severity as well as each other. As there were no results from principal component analysis, it was inferred that the mechanical variables accounted for a similar portion of the perceptual variability. So conflicting perceptions of the event could have occurred.

On the other hand, in some studies (Hennig et al., 1996) surprising negative correlations were found between the first impact of GRF compared to MPF and heel pressures for shoes showing lower first impacts with higher frequencies and heel pressures. This finding was confirmed when correlating the first impact values with the mechanical impactor results. Although vertical force peak was increased with the perception of better cushioning, peak pressures under the heel showed an opposite reaction. These seem to be contradictory results. During rearfoot strike running, midfoot as well as forefoot structures participate in load bearing at the moment of the first vertical GRF peak (Henning, 1995a). Therefore, high in-shoe pressures under the heel would not be necessarily connected with the presence of high peak force values. In a soft-shoe, a higher deformation of the midsole material increases the contact area between foot and shoe. Elevated regions under the arches of the foot experience higher loads on soft midsole materials. As a consequence, the high pressures under such prominent structures like the calcaneus and the metatarsal heads are reduced. Although peak pressures in the heel show higher values for the hard shoe, the relative load results for the forefoot (as given by temporal integration) indicate reduced rearfoot and midfoot loading. This discrepancy can be explained by the difference between the two pressure variables. The relative load analysis incorporates amplitude and duration of the pressure signals. Therefore, in spite of high signal amplitude, a short loading time can result in a reduced mechanical impulse. During rearfoot strike running in the hard shoe the participants have to bear initial weight in the rearfoot, creating high heel pressures. However, the relative load results suggest a rapid unloading of the heel from high impulse loads. This would also explain the reduction in the impact force peak from the soft-shoe to the medium and the hard shoes.

4.4.3. Which footwear properties control impact perception?

Factors such as velocity, surface material, etc. are known to modify biomechanical variables defining impact. However, the relationship between them and impact perception is not clear. Lees (1988) stated that research had been unable to reliably relate perception of cushioning of shoes when running to a linear estimate of the stiffness of the shoe at high physical impact loads nor to resilience. Since running shoe midsole compression is not highly related to perception, damping properties may have greater importance than compliance for eliciting favourable perceptions and for developing intuitively temporal expressions of loading.

Perception has been quite clearly related to biomechanical variables such as input frequency. By contrast, a clear relationship between mechanical properties of shoes and impact perception has not been established. Few studies on perception of cushioning during walking are to be found in literature. On the other hand, the influence of footwear properties on impact perception are likely to be extended to a more general description including foot properties and the modification of both due to passive interaction which will be later described.

Traditional theories postulate that humans perceive an optimal level of hardness and make adjustments to it (Hennig et al., 1996). That was supported by results from a study on running in three shoe constructions in which increased midsole hardness caused a change in foot contact pattern towards increased forefoot loads by reducing lateral rear- and midfoot impulses (Hennig et al., 1996). They found a reduction of the impact force peak with increasing shoe stiffness for three pairs of shoes with comparably large differences in cushioning properties, which was attributed to locomotor adaptation behaviour to avoid high heel impacts. Recent studies have related interface material density, (Lafortune et al., 1998) and stiffness (Milani et al, 1997a, b) to impact perception.

4.4.4. Sensitivity of human perception ability

Some research have postulated that perception ability may be limited in humans (Lafortune et al., 1995b). If so, then in some cases the results obtained for running could not be extrapolated for walking. According to Hennig et al. (1996), it appears that running at a given speed imposes basic kinematic and kinetic demands to accelerate the centre of gravity and footwear can only play a minor influence on these basic biomechanical demands.

Conflicting results with respect to sensitivity of impact severity perception are found in literature. Lake and Lafortune (1997, 1998) found that for the less severe impact conditions in human pendulum test, small changes in conditions were not readily "sensed" by the participants. The authors explained the absence of change in muscular preparation prior to impact in human pendulum testing as being due to the unexpected adjustment in impact severity not perceived by the participants. If the change in impact severity was not perceived then the ability to discern such changes is limited. These findings are coincident with the recently reported low perception abilities of runners wearing commercially available running shoes (Milani et al., 1996). It appears that it was the complex construction features of commercially available running shoes that influenced the runner's recognition of biomechanical variables.

However, the results of a study by Milani et al (1997) showed surprisingly good relationships between biomechancial variables and perceived scores for running in similar shoes with only small differences in midsole hardness in the heel and midfoot area, but identical in the forefoot. Similar results were obtained by Hennig et al. (1996) who reported that in running tests, runners were able to detect the differences between medium and soft-shoes despite little differences between them.

In the work of Lake and Lafortune (1998) participants did not move and only the interface material was changed. Participants were not able to differentiate between impacts on the medium and high-density interface at the slowest impact velocity or between all impact velocity on the low density interface material. For these particular adjustments in initial impact conditions, the changes in the mechanical input variables were relatively small. The difference in material properties between the medium and high-density interface was not as large as the difference between low and medium density. This would explain why changes in impact perception and mechanical input variables were much less between the medium and hard interface. Nevertheless, there were small changes in impact severity that were not perceived differently by the participants. Another factor that may partly contribute to the limited perception sensitivity is the frequency content and conditions that differed from impact experienced. For both impact force and shank shock, the signal power of frequencies above 20 Hz decreased substantially when the high and medium density interfaces were replaced with the low-density interface. There are specific mechanoreceptors that respond preferentially to high frequency mechanical stimuli in excess of about 40 Hz (Griffin, 1990). It can be hypothesised that because these receptors are tuned to high frequencies, it may be more difficult to perceive the severity of impact conditions that generate much lower signal power of above 40 Hz (e.g. all impact conditions on the low-density surface).

At the greater impact velocities, humans are able to perceive differences in the severity of impacts on the medium and hard interface. The main reason for this apparent improved sensitivity is that the differences in impact severity between these interfaces at the higher velocities became substantially larger. Humans beings have more stimuli to allow them to better differentiate between interfaces. Additionally, it can be speculated that as the impact severity moves toward a level that might be considered damaging, then perhaps different sensory receptors come into play to fine tune our severity perception. The human sensors in the plantar skin of the foot called nciceptors could have become increasingly involved as impacts became more severe. However, their involvement appears unlikely for the present range of impact severity because none of the participants reported any pain caused by the impact loads experienced, and this pain is necessary for nociceptors stimulation.

This is very important given that if humans have limited sensory capabilities to detect the magnitude of impact related quantities, body protection by muscular adaptation is restricted and high cushioning levels will lead to decreased perception. These results may have serious implications for walking, since if perception is limited walking impacts could be too low and too slow to be not perceived by the proprioceptive system. In fact, this is not the case as people perceive heel strikes when walking. For high impact forces and velocities that only occur in running lower a threshold of sensitivity has been proposed which would limit perception to running only. However, walking studies of impact perception are not known.

4.5. System Goal

The objective of the proprioceptive system is unclear: *is it to keep any biomechanical variable potentially harmful at a safe level or to improve comfort?. Both have been proposed.*

A number of studies have found that for different accelerations in the femur associated with different walking velocities, no significant differences are observed in the magnitude of acceleration at the head (Voloshin & Wosk, 1981; 1982; Wosk & Voloshin 1981). Based on these observations, it seems that the aim of the system is to protect the head from high accelerations, limiting these to acceptable levels. However, in other studies the magnitude that appears to remain constant is that of the impact force measured by means of force platforms (Misevich & Cavanagh, 1984). If the ultimate aim of the control system is to maintain the level of acceleration in the head, and since soft materials reduce acceleration level in the tibia, then less impact attenuation would be required on the upper part of the human body. In this sense, differences in head acceleration with different walking impact conditions should be investigated. Also, the rate of impact force loading and frequency parameters should be analysed given their relationship with impact perception.

On the other hand, Robbins theory (Robbins et al., 1988a, 1988b, Robbins & Gouw, 1989) is based in plantar discomfort as the mechanism for avoidance behaviour onset. Comfort is described as the feeling that a human perceives of a given product and itself is an objective of footwear users. Luehti and Nigg (1985) and Nigg et al. (1986) analysed comfort and pain in two different tennis shoes. The results showed that almost 40% of humans had experienced some kind of injury related to footwear properties, discomfort being the most frequent (55.1%). **The role of comfort** as an objective of the proprioceptive system, however, has not been studied.

Comfort is a complex phenomenon depending on many factors. It is the result of the combination of aspects such as fitting, shock absorption, pressure distribution, foot movement and others. Nonetheless, comfort as considered in this context excludes thermal comfort. Currently there is controversy about the term comfort (Zhang et al., 1996) since in footwear design, and considering Robbins theory, the concept used is to avoid discomfort through functionality and to avoid injury. Comfort includes more complicated concepts such as user feelings, aesthetics, etc, which are out of the scope of this work. Little research has been devoted to investigate the relationship between comfort and biomechanical variables, or human-shoe characteristics (Luehti and Nigg 1985; Nigg et al. 1986).

Shoe fitting has been described as the main factor determining shoe comfort (Perkins, 1993). Shock absorption and attenuation have been also suggested to play an important role. It has been considered that the sensory faculty of the human body to perceive the severity of impact loads is probably important in the evaluation of comfort (Hennig et al., 1996). Previous results at the IBV showed that there is a significant correlation between the perception of a sole as too hard, overall comfort and, an increase of pain in the heel and lower back.

However, other authors consider that comfort is more related to plantar pressure distribution and its perception according to Robbins theory. Whittle et al. (1997) related the perceived comfort of walking on carpets to their cushioning properties as determined from mechanical impact testing. These authors found that comfort was best related to the response of a carpet to transient forces and argued that pressure distribution across the sole of the foot might be important for the perception of comfort during walking. The importance of plantar pressures was confirmed by Chen et al. (1994) who demonstrated that pressure distribution between the plantar surface of the foot and shoe play an important role in the feeling of comfort in walking shoes. These authors found that peak pressure and pressure-time-integral were more sensitive to the change of the comfort conditions than maximal force and force time integral. In contrast, Jordan and Barlett (1994) reported that differences in perceived plantar comfort between three types of footwear were not related to any in-shoe plantar pressure distribution parameters (including peak pressure and pressure-time-integral). Using a 5-point perception of comfort scale together with plantar and dorsal in-shoe pressure measurements during walking in three different pairs of footwear, the authors concluded that the measurement of foot pressures should be useful for identifying causes of discomfort in footwear. On the other hand, pressure beneath the heel has been also related to impact perception (Hennig et al., 1996; Milani et al., 1997a, b).

Thus it could be the link between comfort and impact perception.

Comfort is a subjective feeling and to analyse it requires subjective techniques. Methodology currently used in comfort analysis was developed 30 years ago in the ergonomics area (Shackel et al., 1969) and has been widely used in different fields of industrial ergonomics such as furniture and workplace analysis (Corlett, 1989), but not in footwear analysis. These methods are based on the Likert scales collecting information about global comfort (Shackel et al., 1969), comfort in different body areas (Corlett and Bishop, 1976) and human's perception on different product aspects (Shackel et al., 1969). The methodology was three different types of questionnaires that were completed before, during and after using a product under controlled conditions.

That humans accommodate to the impact conditions has gained force as the source of differences found in the results between machine and human testing of shock absorbing materials. This behaviour corresponds to a proprioceptive model in which certain mechanical and biomechanical variables act as input to perceive the severity of impact condition. This is matched against a given goal by a control system, which if necessary issues an output signal that - acting on active natural absorbing systems - adjusts input to correct the control variable. This theory is not new but even though extensive research has been devoted to it, little is known about input dependence upon biomechanical dimensions, material properties amongst others. The operation of impact perception and even the methodology to investigate this are not clear. Furthermore, the objective of the system and how the system relates to the output remains to be ascertained.

Thus, more research has still to be done to advance in the knowledge of human shock absorption during walking by investigating the research fields described in the previous Chapters.

The detailed programme of experiments designed to explore and analyse these different fields of research is presented in the remaining Chapters of this thesis.

5

Experimental Design

5. Experimental Design

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The aim of this thesis was to investigate the different statements proposed to explain the differences between human and machine testing of shock absorbing systems. Understanding the origin of these differences is highly relevant for enhancing and advancing in the knowledge base of shock absorption during human walking. Those statements are:

- That current machine testing does not accurately simulate impact forces.
- That materials degrade with use showing inferior properties than measured by machine testing
- That footwear effects on walking kinematics and heel pad confinement are more significant than insole materials in shock absorption
- That humans make accommodation to walking conditions

The objectives of the experiments reported in this thesis were designed to explore these statements. To do so, several hypotheses were established and tested. Null hypotheses were also considered in a way that either new tests were deemed necessary to investigate that statement or it was disregarded as the source of differences. In this sense, the statements were investigated in order of increasing complexity (Figure 5.1), from the straightforward to the complex, and for this a programme of different experiments was designed.

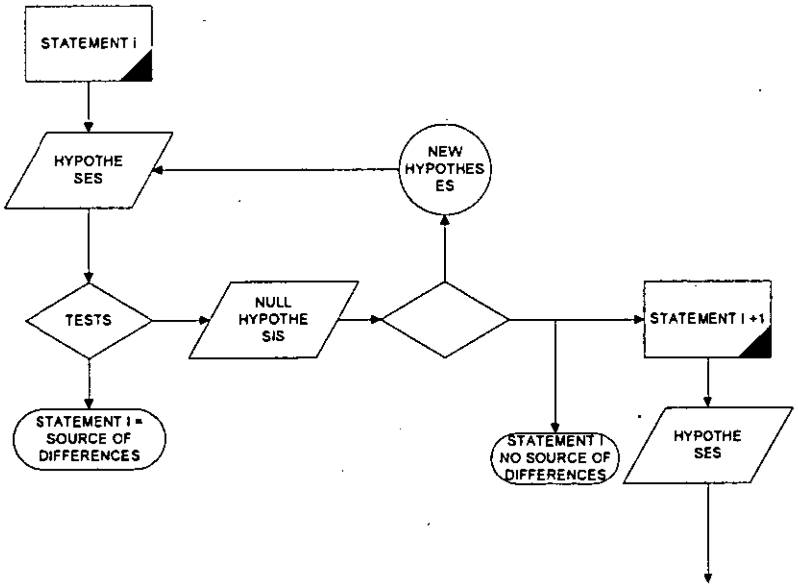

Figure 5.1. Experimental Design

The experimental programme outlined below was designed to address each statement according to the hypothesis and questions to be explored. In the process, it was necessary to explore and solve many biomechanical and methodological issues related to the statements that were investigated.

➊ *Current machine testing does not accurately simulate impact forces.* To explore this statement, it was deemed necessary to test materials simulating impact walking forces and compare the results with walking tests. In this sense, the following hypothesis was tested:

- If materials are tested using a machine method that simulates the magnitude and duration of walking impact forces, a good prediction of material's performance during walking might be obtained. A good agreement would be obtained between mechanical properties of materials obtained by machine testing and ground reaction forces and accelerations during walking with the same materials.

The null hypothesis established was that.

- Testing materials by simulating impact forces is not enough to predict their performance during walking. In consequence this statement would be disregarded as the origin of differences between machine and walking results. So, other statements should be explored.

The work initially attempted to test this hypothesis by using a machine method previously developed at the IBV. This method uses a universal dynamic testing machine to simulate impact loading during walking to measure stiffness and energy absorption of materials. The obtained results were compared with those from walking tests. The mechanical properties of a sample of materials were measured using the IBV machine testing method and compared with forces and accelerations registered in walking tests of the same materials. The work carried out; results, discussion and conclusions are included in Chapter 6.

Experiments for analysing the other statements were designed in a way that maximises the data acquisition with the minimum amount of testing, registering different signals in relation to different statements during the same testing sessions. Later these signals were used for different analysis, as appropriate in relation with the different hypotheses established in each statement.

❷ *Materials degrade with use showing inferior properties than measured by machine testing.* Two questions were addressed in relation to this statement, first the way of measuring material lose of properties and second to relate it to walking test results. Hence, the following hypotheses were considered concerning this statement:

i. That residual displacement of materials measured when a compressive load simulating walking underfoot pressures is removed, is related to long term performance of materials in a way that the greater the residual compressive displacement the more the material degrades with use. In this sense, it was hypothesised that materials showing greater residual displacement would give rise to subjective perception of harder and less comfortable materials due to lose of properties, and
ii. That materials showing greater residual displacement will result in higher peak forces, shorter time to peak force and higher accelerations during walking due to lose of properties.

The null hypotheses were as follows:

i. That degradation of materials in use cannot be predicted by measuring the residual compressive displacement by simulating walking pressures and that further research would be required to check this statement. A good method of characterising lose of material properties with use should be developed, and
ii. That residual displacement as a measure of materials' degradation in use does not account for differences between human and machine testing of shock absorbing systems. As a result, other statements should be investigated.

To test both hypotheses, it was necessary to develop a new machine testing method simulating plantar pressure loading in order to measure the recovery-ability of a sample of materials. That required registering walking plantar pressures with different underfoot materials and people, as well as statistical analyses to define loading conditions and test parameters. The test was then used to obtain the residual displacement of a choice of materials and results compared with those from long term subjective walking experiments to test the first hypothesis and with shock absorption testing during walking in relation to the second. Comparison was done by Pearson's correlation analysis.

The set up procedure for the machine test, choice of materials and preparation for walking tests results was done at the same time for all the experiments designed to explore the other statements. The set up procedures and materials selection are included in Chapter 8 and the walking tests in Chapter 9; the analysis to test both hypotheses is included in Chapter 11.

➌ *The effect of the shoe in walking kinematics and heel pad confinement has greater influence on shock absorption than the materials used.* The difficulty to investigate this statement was double. In first place, it was necessary to separate the effect of the shoe in walking kinematics and heel pad confinement from materials shock absorption which was done by testing different materials into the same shoe (i.e. same kinematics and heel pad confinement) in a way that any measured difference should be due to insole material. Secondly, to differentiate if material's influence was due either to its shock absorbing properties of to other effects. Thus, the hypothesis considered with respect to this statement was.

i. That impact forces and accelerations registered in humans walking tests with different insoles into the same shoe will show no differences. That will confirm low material's effect.

And the null hypothesis:

i. That any differences detected the effect of the insole materials due to either their role as shock absorbers, their influence in human accommodation or to passive interaction. As a result, further research should be conducted to ascertain material's influence.

In consequence, walking tests were conducted with people wearing ten different insoles in the same type of shoe and differences in impact forces and accelerations assessed. The materials' selection process is described in Chapter 8 and human testing in Chapter 9. The results, discussion and conclusions are described in Chapter 11.

Three possibilities arose from the null hypothesis to explain material's effect: accommodation, shock absorption and passive interaction. They were investigated by testing the following hypotheses.

i. *That if the material's effect was to elicit accommodation during walking,* no differences would be observed in impact forces and accelerations registered during passive (no participant's movement) impact testing of people wearing different insoles into the same shoe. Further research should be then conducted in relation to the fourth statement, which refers accommodation as the source of differences.

To test this hypothesis it was necessary to develop a passive impact test to investigate the barefoot heel pad mechanics, which is described in Chapter 7. This method was then used to test a group of people wearing a sample of ten insoles into the same type of shoe. The material selection is included in Chapter 8 whereas passive test on shod people is described in Chapter 10. Again results and analysis are presented in Chapter 11.

The null hypothesis was:

i. That any differences detected in passive tests are the effect of the insole materials due to either their role as shock absorbers or to passive interaction. As a result, further research should be conducted

In this case, it was considered that if differences in passive testing were due to materials' shock absorbing properties, it should be reflected in results from a correlation analysis between the mechanical properties of people's heel pad, material's properties and impact forces and accelerations registered from active and passive tests. In a sense that, the hypothesis to test was that.

i. If material's role was as shock absorber, good correlation should be expected only between impact parameters measured in passive testing and material's properties obtained from machine tests.

The same consideration was made for passive interaction as the source of differences in passive testing. In that case, the hypothesis to test was that.

i. Passive interaction between people and material would be described by a good correlation of passive impact parameters with both material's properties obtained from machine tests and heel pad properties obtained from barefoot pendulum tests.

And the null hypothesis:

i. There was a further effect of material not described yet and further research should be done in this sense.

The materials' selection and properties are described in Chapter 8. Walking tests are included in Chapter 9 and passive tests in Chapters 7 and 10. The correlation analysis, results, discussion and conclusions with respect to the different hypotheses are presented in Chapter 11.

❸ *Accommodation to impact conditions.*

Accommodation is related to human perception of impact forces. The human body has been described as a feed-back system that accommodates to walking conditions relying on impact perception. It is a complex phenomenon. Thus, several hypotheses and questions were considered to explore this statement.

The first question to consider was about whether accommodation really occurs. In this sense, in relation to the previous statement, the hypothesis considered was that accommodation took place if:

i. No differences were observed in impact forces and accelerations registered in passive testing of humans wearing ten different insoles and the same shoe type whereas differences appeared from walking tests with the same people and insoles.

Accommodation was expected to be reflected by the relationship between impact forces and accelerations during walking. Hence, correlation analysis was done and it was considered that

i. Accommodation should be described by any reverse relationship between impact forces and accelerations from walking tests.

The null hypothesis was that accommodation does not happen to introduce differences between machine and walking results.

This work is described in Chapters 9 and 11.

With respect to accommodation, there were many questions to be explored concerning the proprioceptive model. These questions were examined in relation to the different components of the proprioceptive model: sensing, goal, input and output.

Sensing function.

One of the major concerns when studying accommodation was to ascertain whether the ability of humans to perceive impacts is limited and hence modification of impact conditions due to insole change was not perceived by people. The influence of methodological aspects on results was also investigated. The sensing function was explored in relation to impact perception. Three different perception testing methods: two active and one passive, were developed and the following hypothesis considered.

i. People are able to express differences in impact conditions perceived when the insole material is changed both in active and passive testing.

The null hypothesis was that human ability to perceive impacts was limited. Thus, other experimental conditions (other insole materials and/or walking speed, etc.) would be needed to study accommodation.

Goal of the system.

The goal of accommodation has not been described being comfort and health the most outstanding possibilities. With respect to health, it was considered that the goal should be to keep one or more biomechanical variable describing impact forces or accelerations at a safe level when changing insole material. In this sense, the following hypothesis was stated to consider health preserving as goal in case:

i. That no significant differences will be found between insoles for impact forces and accelerations during walking tests.

The same could be considered for comfort as system's goal with the following hypothesis:

i. That there are no differences in comfort level collected with different insole materials.

To test these hypotheses the variability of results from walking tests both objective and subjective with different insole material was analysed.

The null hypothesis was, for both cases, to disregard both as system's goal remarking the need for further research in this topic.

Input

The system's input is the result of interaction between the human body and the ground by means of underfoot materials and the heel pad. In this sense, the input can be studied at two different levels:

i. The biomechanical variables resulting from interaction which are used by humans to sense and evaluate impact conditions, and
ii. The influence of materials' properties, which can be manipulated to "design" impact conditions.

The biomechanical variables considered likely to influence impact perception were those describing underfoot pressures, foot and leg movement and impact forces and accelerations. To explore the first issue, results from walking and passive tests were compared to the results from impact perception testing. Those variables showing a high correlation with impact perception were considered to be used by the system to estimate impact conditions.

The materials properties considered as input for the system were obtained from the following machine tests:

- IBV method for dynamic impact test
- Static compression
- Hardness and density
- Dynamic cushioning.

To explore the influence of materials it was necessary to set up a machine testing method to measure dynamic cushioning simulating underfoot pressures. The material's properties were then compared with results from perception testing. Those properties that correlated high with impact perception were considered to be used as design parameters to control system's input.

The machine testing and the insole materials selection criteria are described in Chapter 8 whilst Chapters 9 and 10 describe human testing. The analysis of the results, discussion and conclusions are presented in Chapter 11.

Output

Output refers to the action performed by the proprioceptive system as response to impact conditions (input). The hypothesis considered was that

i. Accommodation was reflected by a relationship between biomechanical variables describing foot and lower leg movement, plantar pressures and impact forces and accelerations.

The null hypothesis was that further research was required to identify the output.
It was explored by analysing the relationship between biomechanical variables registered to describe plantar pressures, foot and lower leg movement and impact forces and accelerations during walking tests. The human testing is described in Chapter 9 whereas the analysis of the results, discussion and conclusions are presented in Chapter 11.

As already stated, the design of experiments minimised the experimental work in such a way that the machine testing, materials selection, walking and passive tests required to explore the different statements were done concurrently whenever possible. To make this report easier to follow the structure of the thesis is described as follows.
The work conducted to explore the first statement together with results, discussion and conclusions are described in Chapter 6. The ballistic pendulum developed in this thesis for passive testing is described in Chapter 7 along with the analysis of heel pad mechanics and the influence of gender, age and obesity on heel pad properties which was a prerequisite to study heel pad confinement and passive interaction.
The study involved a set of ten insole materials placed into a single model of shoe that was worn by ten male participants. These materials were selected from a larger sample and were representative of the available range in terms of mechanical properties as obtained from the machine tests. The set up of new machine testing methods, machine testing and the selection criteria for a sample of ten insoles for walking and passive tests are described in Chapter 8.
Subjective and objective tests were carried out to check the different hypotheses. The subjective tests included impact perception in both passive and active conditions and comfort testing. The objective tests were done in active and passive conditions. The latter consisted of pendulum testing of barefoot and shod participants. Shock

absorption, plantar pressure distribution and rearfoot movement were analysed in active conditions. Objective tests during walking and subjective test (comfort and impact perception) are presented in Chapter 9. Passive tests of shod humans using the ballistic pendulum are described in Chapter 10.

Finally, Chapter 11 presents the analysis of data collected and results obtained in previous Chapters to explore each statement, to test the hypotheses established and resolve the questions stated with conclusions and recommendations for further work. That chapter presents the discussion for all four statements studied in this thesis, followed by conclusions on each of them and finally conclusions of general interest about walking kinematics, heel pad, etc. and recommendations for further work.

6

Analysis of a new Machine testing method simulating walking impacts to evaluate the efficiency of some materials for external shock absorption

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The Chapter discusses work done to analyse the research question that existing machine tests fail to match the walking test results due to unrealistic machine loading conditions. A new machine test that simulates real walking loading conditions was used to test some insole materials and the results compared with performance in walking tests in order to validate the method. This new test machine was developed at the IBV

6.1. New Material test

A new material test that simulates the loads and rate of loading activities such as running and walking was developed by Garcia et al. at the IBV (1992). With this test both the rigidity and loss tangent (energy loss) are obtained. Rigidity is mainly related to the ability of the material to deform under applied forces (acting as a spring), whereas loss tangent is related to its ability to dissipate part of the energy applied. Rigidity is the ratio of load to displacement while loss tangent is the ratio of energy lost to energy stored in one stress-strain cycle applied to the material. These parameters are not necessarily correlated as, for example, a quite soft material can be non-energy absorbing (elastic), while contrarily, a quite rigid material can be energy absorbing (viscoelastic materials in certain circumstances).

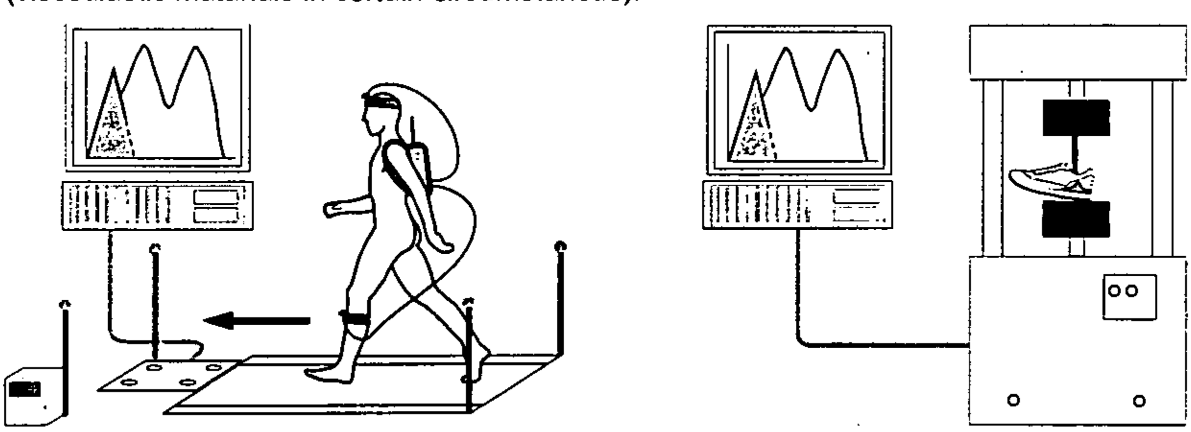


Figure 6.1. IBV method for machine testing of footwear materials

Materials testing was accomplished using a computer controlled Instron (model 8501) dynamic-testing machine at IBV's laboratory. The load and displacement transducers in the machine produce analogue electric output signals that are used to compute

rigidity and loss tangent through a double channel spectrum analyser that performs frequency analysis of both (García et al., 1994).
The complex impedance of materials is used to obtain rigidity and loss tangent. The complex impedance of a given linear material is defined as $G(\omega)= \sigma(\omega)/\varepsilon(\omega)$, where σ is the stress and ε the strain and ω is frequency. For a harmonic stress excitation of frequency equal to ω, and the stress given by $\sigma(\omega)= \sigma_0 \sin(\omega t)$, the steady-state strain is: $\varepsilon(\omega)= \varepsilon_0 \sin(\omega t - \delta(\omega))$. $\delta(\omega)$ is the phase shift angle.
In complex representation,

$$\begin{aligned} G(\omega) &= \sigma_0(\omega)\varepsilon_0(\omega)e\, j\delta(\omega) \\ &= G1 + jG2 \\ &= \sigma_0(\omega)/\ \varepsilon_0(\omega) \cos \delta(\omega) + j\sigma_0(\omega)/\varepsilon_0(\omega)\sin\delta(\omega) \end{aligned}$$

In this expression, the real part is related to the stored energy per cycle and the imaginary part to the lost energy per cycle. The ratio of the imaginary part to the real part is equal to the loss tangent and the modulus of the complex impedance gives the rigidity (stiffness) of the material.
Fast Fourier Transform is used to calculate the complex impedance of materials as a function of frequency. Data in dB for all loading frequencies is obtained from the spectrum analyser and converted to System International Units using Excel by Microsoft. Only frequencies below 50 Hz are normally considered since coherence becomes less than one with greater frequencies.

6.2. Methodology

In a previous work several materials were studied with the new method (García et al., 1994), simulating walking forces. These materials include PORON 5, NOENE and IMPLUS. As a result of these studies it was found that NOENE was the most energy absorbing material, while PORON 5 had the lowest rigidity. IMPLUS was found to be comparable to NOENE with respect to rigidity and comparable to PORON with respect to loss tangent. Garcia's work was confined to machine testing and it was important to compare these results with human testing. The above materials were selected to study the correlation with human testing because their mechanical characteristics with respect to impact absorption - rigidity and loss tangent - presented extreme values. Figure 6.2 shows the loss tangent and rigidity of the three materials.

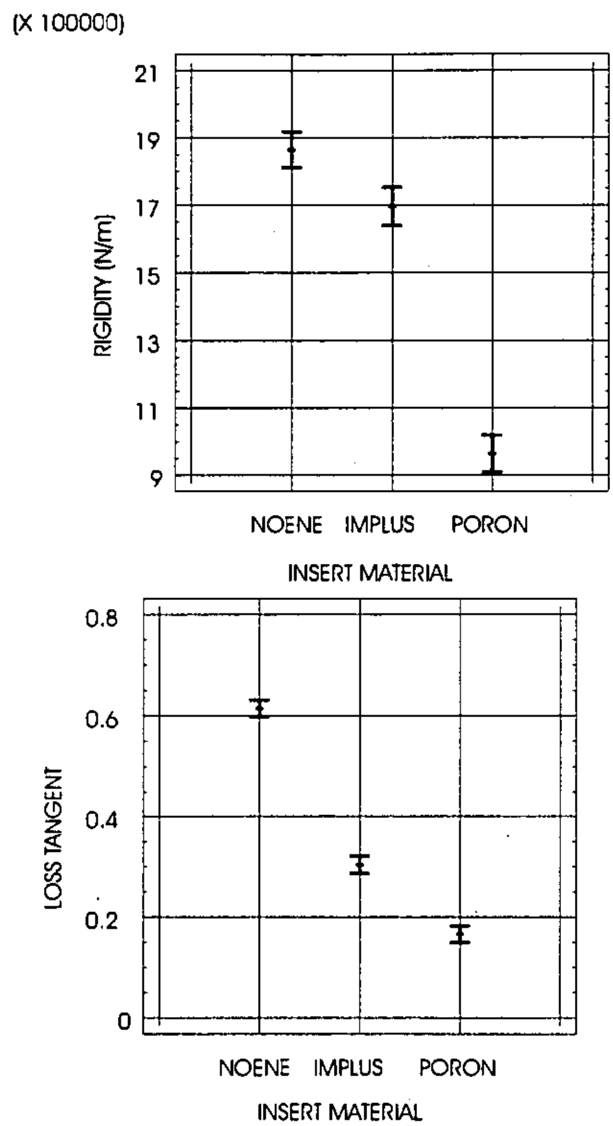


Figure 6.2: Results of the material test performed under walking forces. a) Rigidity. b) Loss tengent

To check the hypothesis that if walking loads simulated by machine testing gave good predictions of performance of materials, then the performance of these materials during walking could be studied in detail. Insole inserts were made from each material and inserted into the shoes. The objective was an improved understanding of the relationship between the mechanical characteristics of the inserts and effectiveness in cushioning during walking. This would be achieved by examining the influence of two mechanical properties of the materials (rigidity and loss tangent) on the generation of impact forces and the transmission of shock waves through the musculoskeletal system; this would also establish criteria for future selection of insert materials.

Five healthy males were selected for the tests. All participants were under 35 years of age with active life-styles and who walk habitually; all gave informed consent to be part of this study.

Insoles with inserts (Figure 6.3) of IMPLUS, NOENE and PORON 5 were specially made for the study. IMPLUS (Texon, USM, Leicester, England) is a PVC foam of an open-cell structure. NOENE is a material with a basis of a micro-air elastomer first developed to isolate machinery from vibration. PORON (Rogers Corp., East Woodstock, CT) is a cellular urethane available in various grades. A grade commonly used in footwear applications was selected. As already stated, these materials were selected because their mechanical characteristics as they relate to impact absorption (rigidity and loss tangent) represent extreme values.

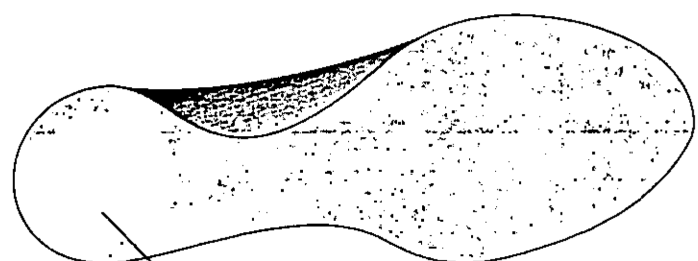


Figure 6.3. Insole with viscoelastic heel insert

Insoles were tested when worn in a leather street shoe. Only one type of shoe was chosen for experiments to eliminate footwear effect in heel pad confinement and walking kinematics so that any differences recorded would be due only to the insert material.

To measure the transmission of the impact wave through the body, two accelerometers were used, one placed on the lower limb (Figure 6.4) and the another on the head. Accelerometer attachment on the skeleton fulfilled the requirements mentioned previously (Introduction). Lower limb placement was chosen as the proximal anterior part of the tibia, 3-4 cm under the tibia tuberosity, on the internal part. The sensitive axis of the uniaxial accelerometer was chosen to be in the same direction that the longitudinal axis of the tibia. The accelerometer was an ICSENSORS 3031. The performance specifications supplied by the manufacturer of this piezo-resistive transducer are: range 20 g, resonance frequency 1200 Hz, sensitivity 2.1 mVg^{-1}, weight 0.3 grams.

Figure 6.4. Accelerometer placed in the lower limb

The accelerometer was attached to the skin by double side adhesive tape and an aluminium support. The weight of the system was less than 2.5 g, i.e., well below the critical limits reported in the literature to measure bone acceleration from skin mounted accelerometers (Lewis et al., 1991; Ziegert & Lewis, 1979). A bandage wrapped tightly around the shank was used to fasten the accelerometer and to pre-load the skin.

For the accelerometer location on the head an earlier experiment was performed. A bite-bar (Light et al., 1980) and a forehead fixing similar to the one used on the tibia were tested. The two signals were very similar but the forehead fixing had greater repeatability and participants were more tolerant of this kind of fixing. Moreover, the forehead and tibia fixing are similar and, hence, the transmission factor computed as the ratio between the signals from the tibia and head is less affected by the type of the mounting system (Smeathers, 1989). For these reasons, forehead fixing was chosen and the same accelerometers and supporting devices that were used on the tibia. A similar elastic bandage wrapped these devices around the head. The signals from both accelerometers were linked to a telemetry system on the back of the participant, in the form of a rucksack, and a receiving system connected to a computer. Accelerometer attachments and laboratory configuration is shown in Figure 6.5.

Each participant was asked to walk with the attached accelerometers and step with the right foot on the force plate (also developed at the IBV (Dynascan®)). Force plate and accelerometers were synchronised by triggering the start of data acquisition via a photocell-emitted pulse when the participant crossed the photo-barrier. Force plate and accelerometer sampling rate was 1000Hz. The sampled signals of both force plate and accelerometers were stored for further analysis.

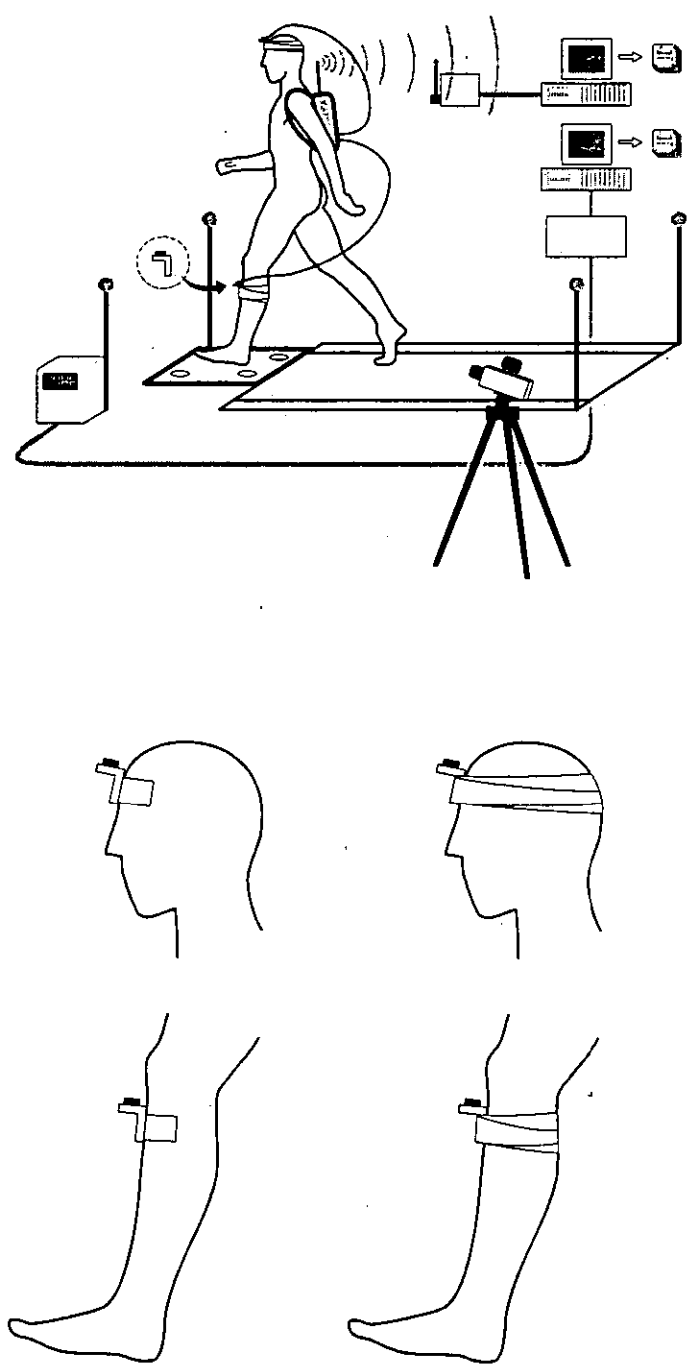


Figure 6.5: Configuration of the experimental set-up. a) laboratory configuration. b) accelerometers attachment.

To control the effect of velocity, the walking rate was fixed at 2 ms^{-1} and trials were considered valid if the recorded velocity was within 10%. The velocities were measured by means of two pairs of photocells connected to a chronometer. For accommodating trials it was necessary for the participant to step on the force plate at a steady rate and without change of movement pattern. The effect of leg inclination at heel strike in tibia acceleration was neglected. Participants were video recorded as they stepped onto the force plate and if changes in either the movement pattern or head inclination were observed the trial was discarded. Each participant performed 10 valid trials in the same shoe under each of the following conditions:

- NOENE insert in the insole inside the shoes
- PORON 5 insert in the insole inside the shoes
- IMPLUS insert in the insole inside the shoes
- Barefoot

Conditions were randomised for each participant to minimise any order effect. To minimise the effect of shoe type on the results the participants wore the same shoes for all the trials with only the inserts changing. The shoes were standard laced men's leather dress shoes with a leather sole and a layer of rubber (80 Shore A) in the heel. The signals of both the force plate and the accelerometers were stored and several parameters in time domain were subsequently computed. These parameters were:

Force parameters (Figure 6.6)

- Initial maximum vertical impact force (Fi)
- Time from touchdown to initial maximum vertical impact force (tFi)
- Rate of loading from contact to the time of initial maximum vertical impact (Fi/tFi).

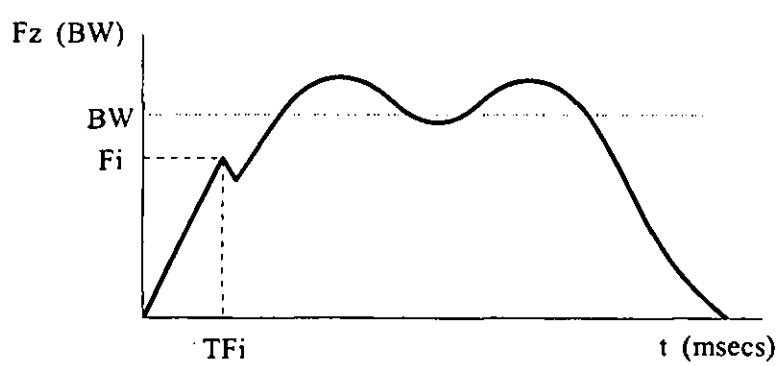


Figura 6.6. Example of typical signal of ground raaction force showing paramaters extracted for statistical analysis.

Acceleration parameters (Figure 6.7 a, b)

- Maximum acceleration on tibia (ati)
- Time from minimum to maximum acceleration on tibia (tati)
- Rate of acceleration on tibia equal to ati/tati

- Maximum acceleration on forehead (afi)
- Time from minimum to maximum acceleration on forehead (tafi)
- Rate of acceleration on forehead equal to afi/tafi
- Ratio of maximum acceleration on forehead with respect to maximum acceleration on tibia equal to afi/ati.

Acceleration versus force parameters

- Ratio of maximum acceleration of tibia with respect to initial maximum vertical impact force (ati/Fi)
- Ratio of maximum acceleration of forehead with respect to initial maximum vertical impact force (afi/Fi)

It is widely acknowledged that, in general, lower impact forces, longer impact times and lower rates of loading are considered less damaging. Besides, lower transmission of impact waves is more desirable as it means greater protection of upper structures. Therefore, lower ratios of tibia and forehead acceleration to impact force and lower transmission of impact wave from tibia to forehead are preferred (McMahon et al., 1987; Seireg & Gerath, 1975; Serink et al., 1977; Simon et al., 1981; Voloshin & Wosk, 1981; 1982; Wosk & Voloshin, 1981; 1985;).

All the force parameters are expressed as units of body weight [BW] while the acceleration parameters are expressed as units of gravity acceleration. Times are expressed in milliseconds.

A multifactor analysis of variance (ANOVA) was performed with these parameters with participant and walking condition (barefoot and shod with insoles of the selected materials) as factors. An analysis of variance was also performed with participant, rigidity and loss tangent as factors. Normality and homogeneity of variance were checked by Leben and K-s tests, respectively. The variables in the study are probably not independent. However, the interest of the study was to study the role of material's properties that can be measured by machne testing with a clear meanin rather than in obtaining new vairables by combination of the others, thus Manova was not used.

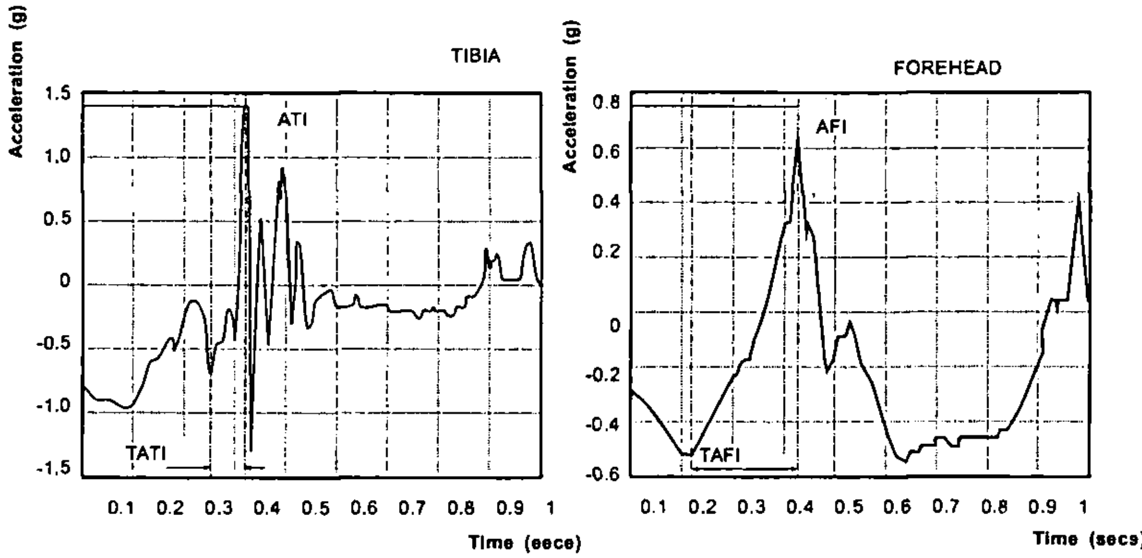


Figure 6.7: Typical signals of tibia and head acceleration, and parameters used for the statistical analysis (description in the text). a) Tibia acceleretion: b) Head Acceleration peremeters.

The levels of rigidity and loss tangent of the materials were from previous walking tests and are as follows:

- NOENE: high rigidity and high loss tangent
- PORON 5: low rigidity and low loss tangent
- IMPLUS: high rigidity and low loss tangent

This method makes it possible to investigate not only the effect of the materials alone but also the separate effects of rigidity and loss tangent to establish criteria for the selection of materials based on their mechanical characteristics.

Significant differences between the different levels of factors were investigated using an LSD (Least Squares Differences) post hoc analysis. An a priori alpha level of 0.05 was chosen. A test power study was done for all the variables investigated to evaluate the probability of finding statistically significant differences of a given size.

6.3. Results

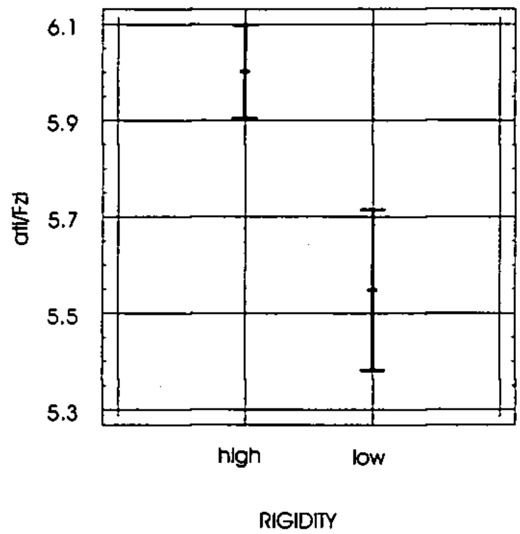
Table 6.1 shows the results obtained for the parameters considered, presented both as a function of material and as a function of rigidity and loss tangent.

A higher loss tangent was found to significantly reduce the ratio of maximum tibia and forehead accelerations to maximum vertical force (ati/Fi and afi/Fi, respectively) (Fig. 6.8.b). This means that for the same level of force, the level of acceleration reaching the body will be lower for materials with a higher loss tangent. It was also found to significantly increase tFi and to reduce the rate of acceleration at the forehead (afi/tafi).

Table 6.1: Results of the parameters studied as a function of meteriels, rigidity end loss tangent. The values are mean end standerd deviations for all the particiepnts and trials. Asterisks indicate significant differences. For configuration as factor one esterisk indicates significant differences between shod end unshod conditions while two asterisks indicates significant differences between materials and shod and unshod conditions. One asterisk between brackets indicates differences between materials not found between shod end unshod conditions. Obtained p-velues ere elso included.

	RIGIDITY (Nm^{-1})			LOSS TANGENT			CONOITIONS				
	High	Low	P	High	Low	P	Barefoot	Noene	Poron	Implus	P
Fi	0.432±0.004	0.445±0.007	0.16	0.445±0.007	0.431±0.004	0.18	0.82±0.001*	0.44±0.01	0.44±0.01	0.43±0.01	0.00
TFi	17.3±0.5	18.9±0.6	0.09	19.1±0.6*	16.9±0.5	0.02	9.4±0.6**	18.4±0.8	17.7±0.6	18.2±0.6	0.00
Fi/TFi	0.0265±0.0006	0.025±0.001	0.23	0.025±0.0001	0.0267±0.0006	0.12	0.094±0.002*	0.026±0.002	0.026±0.002	0.027±0.002	0.00
Ati	2.57±0.04	2.47±0.06	0.36	2.46±0.06	2.58±0.04	0.21	4.46±0.06*	2.51±0.08	2.53±0.06	2.64±0.06	0.00
TATi	63±2*	*54±3	0.02	55±3	62±2	0.08	65±2(*)	60±2	57±2	66±2	0.01
ATi/TATi	0.043±0.001	0.045±0.002	0.26	0.044±0.002	0.442±0.001	0.93	0.060±0.002**	0.043±0.002	0.046±0.002	0.043±0.002	0.00
AFi	1.50±0.01	1.51±0.02	0.49	1.48±0.02	1.52±0.01	0.08	1.6±0.02**	1.48±0.02	1.53±0.02	1.51±0.02	0.00
TAFi	193±5	193±8	0.94	191±8	195±5	0.84	204±6	191±8	195±8	195±6	0.59
AFi/TAFi	0.0078±0.0002	0.0073±0.0004	0.36	0.0071±0.0004*	0.0081±0.0002	0.03	0.0065±0.0003(*)	0.0074±0.0003	0.0079±0.0003	0.0084±0.0003	0.09
ATi/Afi	1.72±0.02*	1.83±0.04	0.05	1.65±0.04	1.71±0.02	0.18	2.64±0.09*	1.69±0.1	1.67±0.09	1.75±0.09	0.00
ATi/Fi	6.0±0.1*	5.8±0.1	0.001	5.5±0.1*	6.0±0.1	0.000	5.53±0.09**	5.74±0.1	5.62±0.09	6.3±0.9	0.00
AFi/Fi	3.53±0.04	3.42±0.07	0.17	3.38±0.07*	3.59±0.04	0.005	2.28±0.06*	3.41±0.06	3.53±0.06	3.64±0.06	0.00

A significantly lower ratio of tibia acceleration to impact forces (ati/Fi) was found for lower rigidities (Fig. 6.8.a). Nevertheless, higher rigidities were found to significantly reduce the rate of maximum acceleration on the forehead with respect to maximum acceleration on the tibia (afi/ati) (Fig. 6.9). Higher rigidities were also found to increase the time to maximum acceleration on the tibia (tati).



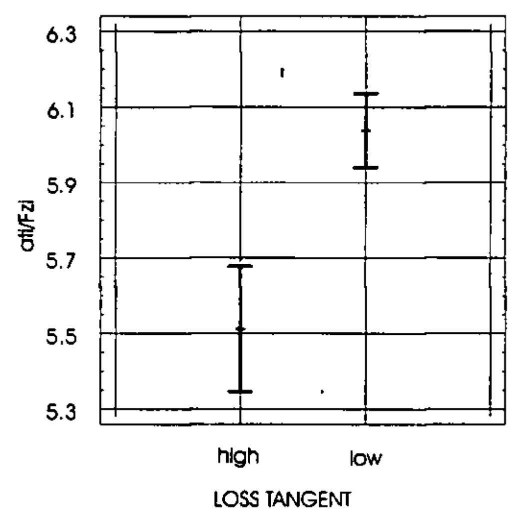


Figure 6.8: Transmission of impacts to tibia. a) As a function of rigidity. b) As a function of loss tangent.

Regarding the test conditions, the most significant differences were found between the shod and unshod conditions. The highest peaks in forces and accelerations were found, as expected, in the barefoot condition. Nevertheless, in time to peak force there were differences between shod and unshod conditions only in time to initial maximum vertical impact force (tFi), while the time to peak accelerations showed no difference walking barefoot or with shoes.

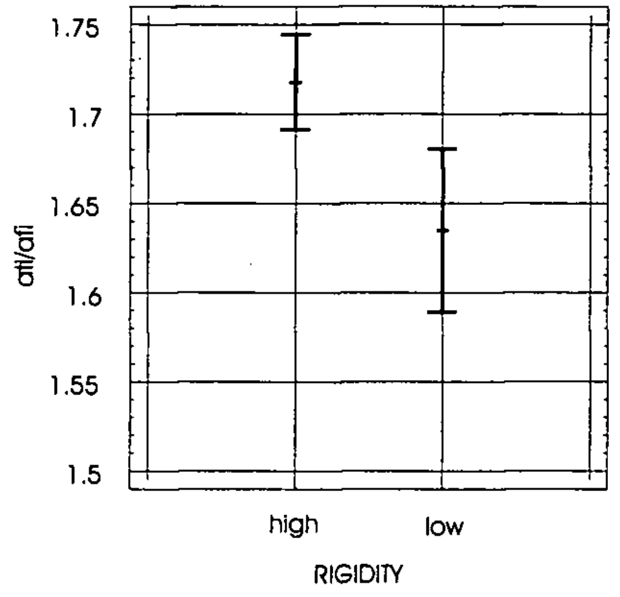


Figure 6.9: Transmission of impacts from tibia to forehead as a function of rigidity.

As a result, higher force rates and tibial acceleration rates were found (Fi/tFi and ati/tati), but no differences in forehead acceleration rates (afi/tafi) were encountered. Transmission of impact to tibia and forehead, expressed by the acceleration divided by impact force (ati/Fi and afi/Fi), was found to be lower for the barefoot condition.

As to the force parameters, significant differences were found between NOENE and IMPLUS, NOENE being the material that showed the highest times of impact (tFi). In the case of tibia acceleration parameters, significantly lower times (tati) of impact were found for PORON 5 and NOENE with respect to IMPLUS. For the forehead acceleration parameters, NOENE was found to significantly reduce forehead acceleration (afi) in relation to PORON 5 and IMPLUS (Fig. 6.10). The acceleration rate (afi/tafi) was also found to be significantly lower for NOENE than for IMPLUS. NOENE and PORON 5 were found to significantly reduce the ratio of tibial and forehead acceleration to vertical impact force (ati/Fi and afi/Fi, respectively) with respect to IMPLUS.

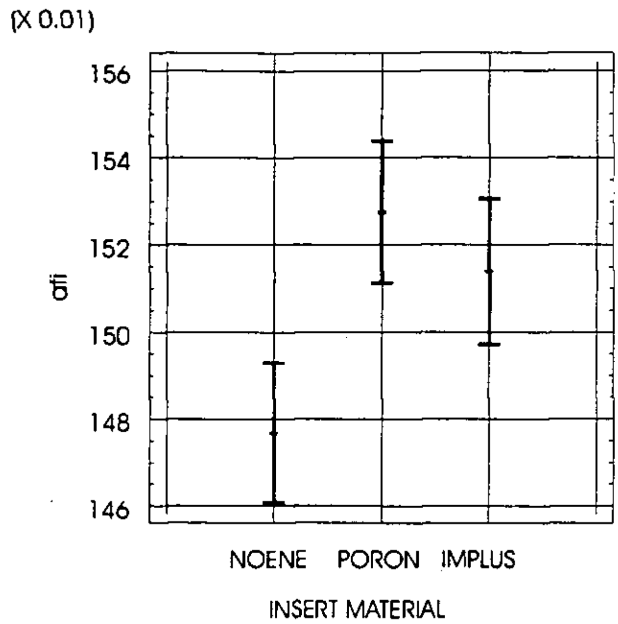


Figure 6.10: Peak acceleration on the forehead in g's as a function of insert material.

6.4. Discussion

The force and acceleration parameters are broadly similar to those reported by other investigators despite the apparent differences due to important experimental factors as walking speed, shoe type, hardness of walking surface or human variability. The values obtained in peak tibial acceleration in this study ranged from 4.48g in barefoot condition to 2.51g with Noene insert, both mean values (Table 6.1). The peak forehead acceleration ranged from 1.8g in barefoot condition to 1.48g with Noene insert. All of these values are well within the range of values observed by several investigators. Light et al (1980) found values between 2g to 8g in peak tibial acceleration, levels of teeth acceleration less than 1g. Nigg & Herzog (1994) present a summary of values of peak tibial acceleration in walking condition from various authors ranging from 2g to 5g.

Force parameter values were similar to those reported by several authors. In the most recent comparison (Shiba et al., 1995) the mean values for shod conditions with different inserts ranged from 31.4% of Body Weight (BW) to 35.1% BW. The results presented in Table 6.1, range from 82% BW (barefoot) to 43% BW (Noene insert). Nigg & Herzog (1994) presents a summary of values obtained in external vertical impact force under different conditions and for various activities. In the walking condition at 1.3 ms^{-1} the values ranged from 55% BW for the unshod condition to the 27% BW for street shoes.

In the present study statistically significant differences have been found between the materials in terms of heel strike transmission to the body. The choice of materials enabled the differentiation of effect attributable to loss tangent from that attributable to rigidity both in transmission and in adaptation to impact. As expected, both low rigidity and a high loss tangent imply a decrease in the impact wave transmitted to the tibia. The mechanical explanation for this is based on the fact that for a given energy input to the system, a less rigid material will reduce the forces by absorbing the energy through elastic deformation and by increasing the time taken to transfer the momentum. On the other hand, a material possessing a high loss tangent dissipates part of this same input energy and as a result the energy absorbed and, therefore, the force involved is less. Thus, a similar effect is achieved by means of different physical mechanisms: a decrease in impact transmission to the body. However, there were unexpected results for shock transmission to forehead. A higher loss tangent reduced forehead transmission as well as tibial shock. But higher forehead transmission was found for lower material stiffnesses.

This good correlation found between the mechanical tests and the trials performed with human beings is not frequently referred to in the literature (Lees, 1988; Robbins and Gouw, 1990; Shiba et al., 1995). The results of this study show that both rigidity and loss tangent (obtained using this new method) exert a significant effect on impact transmission. This means that if only one of these two parameters is measured unexpected effects may result. For example, Noene was found to possess an elevated rigidity and a high loss tangent, while Implus showed a rigidity similar to Noene but lower loss tangent. If only rigidity had been measured the differences in impact transmission recorded between Noene and Implus would have been impossible to explain. As two possible physical mechanisms co-exist to reduce impact force (deformation increase and energy dissipation) testing should evaluate both the mechanical characteristics (rigidity and loss tangent) of the materials related to these mechanisms.

There was good correlation between parameters obtained using the new machine method for transmission to tibia, but not for transmission to forehead through the upper body. Theory of impact perception and proprioception as a mechanism of regulating walking strategy for active natural shock absorption could account for these results.

The forehead to tibia ratio of acceleration (afi/ati) was found to be significantly lower for the unshod conditions, while transmission to the forehead (afi/Fi) was not found to be significantly different for shod and unshod conditions. This fact may be attributable to increased proprioception of impact that activates natural adaptations to reduce transmission to the upper body in the harder conditions. This also produces lower accelerations in the forehead with respect to vertical force. This result is particularly important as for supporting accommodation to walking conditions. It should be noted that no differences were found in time to maximum acceleration between shod and unshod conditions and that with higher rigidities higher times of tibia impact were found. This could imply that natural shock absorption mechanisms activated by proprioception might also increase times of maximum acceleration. Notwithstanding the above considerations, the human body is not a passive system; it possesses mechanisms that reduce the impacts of walking or running. For this reason a perfect correlation between human and material testing is not feasible, as this aspect of the system is not measured by the material test.

As pointed out by Robbins & Gouw (1991) and previously described in detail, the human body may be described as a feedback control system. In human testing the individual adapts to the circumstances of locomotion (Clarke et al., 1983a, 1983b). These circumstances are determined by conditions that include displacement velocity, type of ground and footwear. The interaction between humans and the circumstances of locomotion may be described by means of a proprioceptive model, whereby these circumstances give rise to an input that is perceived by a feedback control system that compares the signal with its potential effects and consequently generates a response to mitigate these effects. Examples of mechanisms for reducing impact include knee bending, contralateral flexion of the hip, and the position of the foot at the moment of heel strike (McMahon et al., 1987; Robbins et al., 1988).

The impact perception mechanism and the functioning of the feedback control system are not clearly understood and comprise part of the goals of this thesis. To investigate how the system adapts to reduce impacts, the relationship between system input and output must be studied. As stated in this Chapter, differences in impact transmission from tibia to head were found to be a function of the rigidity of the material under the human heel. This could mean that the configuration of the kinematic chain was modified on changing the material. The lowest impact transmission corresponded to the barefoot situation, while impact transmission from the tibia to the head was greater for the softest material than for the most rigid. This may be explained as a different adaptation due to different perceptions of impact. A soft material may mask impact perception and hence reduce adaptation. If regulation of the control system depends on perception at insole level, then it is reasonable to suppose that less rigid materials in direct contact with the plantar surface will produce lesser pressures and thus reduce impact proprioception. In contrast, if the ultimate aim of the control system is to maintain the magnitude of acceleration in the head at a safe level then less impact attenuation would be required on the upper part of the body with soft materials that reduce acceleration level in the tibia. However, significant differences were observed in acceleration in the head for the materials studied as the most rigid and energy dissipating material yielded the lowest acceleration in the head. This seems to contradict the system objective to maintain the level of acceleration in the head. Further research is required to clarify this aspect.

A better understanding is required of the perception and adaptation mechanisms of heel impacts. Studies using accelerometers on the tibia and the head and kinematic analyses in running have shown that greater knee flexion give rise to lesser head/tibia acceleration ratios (McMahon et al., 1987). It is reasonable to suppose that a similar phenomenon occurs in walking. Thus, the results of the present study could, for example, explain the cause of greater knee flexion during walking. However, as pointed out above, knee flexion is not the only kinematic adaptation mechanism for reducing the level of heel impact. The use of pressure measuring systems in footwear, and the performing of kinematic studies are required to secure a better understanding of the system of adaptation to heel strikes.

These results suggest that human-footwear interaction occurs and this determines upper body transmission. However, interaction seems to depend on the properties of the insert material since the same shoe was worn in all cases. Forehead transmission appeared to be inversely related to material stiffness that controls impact perception. It seems that impact perception actuates active mechanisms such as knee flexion to reduce shock transmission through the upper body that could be called active interaction. On the other hand, dependence of impact perception on insert stiffness could be extended to global stiffness between the ground and the calcaneus bone, including the heel pad properties. In this sense, a passive interaction between humans and shoe could occur altering impact perception by combination of heel pad and insert properties. The heel would sink more in softer materials and this would increase confinement as the insole material embraces the heel. At the same time, the mechanical coupling of heel pad and underfoot materials is likely to change the material's structure and also mechanical behaviour under initial loading. Again, further research would be needed to more fully understand role of active and passive interaction in impact perception.

Good impact absorption is associated with compliant materials whereas rigid materials are known to increase the efficiency of locomotion: so it could be thought that excessive impact absorption would reduce the efficiency of walking. On the other hand, optimum properties could be postulated to issue good shock absorption while keeping the efficiency of locomotion (McMahon & Greene, 1979; Nigg et al., 1995b). Further research is needed to better understand the influence of mechanical parameters of rigidity and loss tangent in locomotion efficiency.

The magnitude of the differences between materials is about 10%. Some authors point out a relationship between some injuries or degenerative disease and high levels of impact forces and accelerations (Voloshin & Wosk, 1981; Wosk & Voloshin, 1981). However further research is required to establish the levels of forces and accelerations that could have pathological effects.

6.5. Conclusions

Experimental results suggest that the new IBV machine method is capable of providing good correlation with human testing with respect to impact force and tibia shock reduction and confirm that simulating real loading conditions is necessary for testing footwear materials and for a good prediction of material's performance. As expected, lower rigidity and higher loss tangent reduced the transmission of impacts to tibia, although a lowering of the transmission of acceleration from tibia to forehead was found with the most rigid materials. Thus, although a high loss tangent has been observed to reduce impacts, lower rigidity may lead to a higher transmission of the impacts to forehead. As mentioned, this effect can be due to decreased impact proprioception that leads to a lower activation of natural shock absorbing mechanisms such as knee flexing or muscle stress changes.

In general, these results suggest that materials with a high loss tangent and moderately low rigidity should be preferred for reducing impacts transmitted from ground to the tibia and forehead. Optimum rigidity would yield diminished transmission to tibia without lowering impact proprioception and, therefore, without increasing the transmission of impact from tibia to forehead. These results would only apply for people with a normal proprioception and without impaired shock attenuation systems. For those people with impaired shock attenuation systems less rigid materials will be preferred, as they could not adapt to perceived high impacts. For some patients (neuropathic diabetes, metatarsalgia, etc.) it would be more important to avoid peaks of plantar pressure by means of a less rigid material.

Of the materials studied, NOENE best satisfied these characteristics; it was found to significantly reduce impacts in the forehead when compared with both IMPLUS and PORON 5. NOENE also showed significantly lower impact acceleration relative to force than IMPLUS both in the tibia and forehead. Besides, NOENE increased the rise time of vertical ground reaction force and tibia acceleration and reduced the rate of forehead acceleration compared with IMPLUS. No significant differences were found between PORON 5, NOENE or IMPLUS for any of the variables studied, except forehead acceleration that was significantly lower for NOENE. The tibia acceleration to force ratio was also found to be significantly lower for PORON 5 with respect to IMPLUS. Nevertheless, rise time of tibia acceleration was higher for IMPLUS than for PORON 5.

Further research would be required to confirm these results. Relationships between material stiffness, impact perception and shock transmission should be investigated in the context of a proprioceptive model for shock absorption. On the other hand, a deeper analysis of the role of passive and active natural shock absorbing systems would lead to a better understand of proprioception and humans-shoe interaction. In this context, the mechanical analysis of heel pad under impact loading and its coupling with materials underfoot would justify special attention.

APPENDIX 6.A1. POWER OF TEST

A study of the power of the test was done for all the variables, with both configuration and material characteristics as factors, to evaluate the probability of finding statistically significant differences. Due to the very conservative way of computing test power, an 80% probability of establishing an effect of a given size is usually considered reasonable. Table 6.A1 shows the effect size for an 80% probability of establishing significant (alpha=0.05) results for all the variables.

The asterisk shows variables for which significant effects have been found in each study. For configuration as factor, one asterisk indicates significant differences between shod and unshod conditions, while two asterisks indicate statistically significant differences between materials and between shod and unshod conditions. One asterisk in brackets indicates no differences between materials whether shod and unshod. If no significant effect is found for a given variable, then any possible effect should be lower than the values shown in Table 6.A1.

Table 6.A1: Results of test power for all the variables studied.

	CONFIGURATION		LOSS TANGENT AND RIGIDITY	
	Magnitude	% of signal	Magnitude	% of signal
Fi/weight	0.044 *	8.6	0.023	5
TFi	3 * *	19	2 *	11
Fi/(TFi · weight)	0.008 *	20	0.003	11
Ati	0.17 *	6	0.2	8
Tati	11 (*)	18	10	17
Ati/Tati	0.011 * *	22	0.0068	15
Afi	0.06 * *	4	0.005	3.6
Tafi	26	13	25	13
Afi/Tafi	0.0015 *	19	0.0013 *	16
Ati/Afi	0.44 *	20	0.12 *	7
Ati · weight/Afi	0.27 * *	4.6	0.37 *	6.2
Afi · weight/Afi	0.2 *	6	0.21 *	6

7

Heel pad mechanics under walking impact conditions. Principal components analysis and influence of individual characteristics

7. Heel pad mechanics under walking impact conditions. Principal components analysis and influence of individual characteristics

7. Heel pad mechanics under walking impact conditions. Principal components analysis and influence of individual characteristics

This Chapter analyses heel pad mechanics and its dependence on characteristics of the individual. This work was carried out to test the third statement that the shoe effect on heel pad confinement and walking kinematics is of greater influence on shock absorption than the properties of insole material. Although the literature abounds with work on heel pad testing, existing knowledge was insufficient to assess heel pad function under walking impacts and a detailed analysis had to be conducted. A literature survey of heel pad testing is included to indicate current knowledge and to identify inadequacies in present understanding.

7.1. Introduction

The heel pad is the body's most important natural shock absorbing system [Chapter 2]. Its properties have already been described [Chapter 2] and the manner and extent to which they may be modified by confinement due to the footwear, was thought to be the reason for differences between machine and human testing. Hence, knowledge of the heel pad mechanical behaviour is very important for better understanding of shock absorption during human walking.

An analysis of some previous work on basic heel pad mechanics (Denoth & Nigg, 1981; Nigg & Denoth, 1980), Cavanagh et al (1984) concluded that to advance the study of shock absorption during running, the following were required:

i. Reliable and repeatable method to measure mechanical properties of the heel pad in loading conditions similar to running.
ii. Means of estimating the energy absorbing ability of the heel pad.
iii. Assessment of inter-individual variability.
iv. Study of the effect of repetitive impact on stiffness and energy absorption of heel pad.
v. Assessment of the pattern of heel pad deformation during the impact.
vi. Mechanical model to describe these components.

Since 1984, much research has been devoted to the analysis of heel pad properties. However, there is a significant lack of knowledge of heel pad impact mechanics and its dependence on individual's characteristics which makes the analysis of shoe effect on its properties difficult. Research has been mainly focused on running and data on heel pad functions during walking are scarce.

A literature survey reflected confusion. Whilst the heel pad, its general function and properties are basically known, many different testing procedures and a great variety of parameters have been used to describe heel pad properties with conflicting, non-comparable results. Many factors that influence heel pad properties have been identified but no clear relationship has been established which further complicates the understanding of shock absorption.

Different authors have observed common features in the force-displacement curves of the heel pad that reflect a non-linear and viscoelastic nature under impact loading (Aerts et al., 1995; Cavanagh et al., 1984b; Kinoshita et al., 1993a; Valiant, 1984). A great variety of methods and parameters have been used to describe heel pad behaviour under impact loading, many based on static testing methods (Jorgensen et al., 1989b; Prichasuck et al., 1994a; Prichasuck, 1994b). The results obtained during static testing, however, are unrealistic since the mechanical response of viscoelastic materials - such as the heel pad - depends on loading rate. Dynamic testing, therefore, is required but few works using dynamic testing methods have focused on walking conditions (Kinoshita, 1993a, b, 1996a, b).

The literature contains a number of parameters used to describe heel pad properties. Some, such as the visual compressibility index and thickness (Jorgensen, 1985; Khuns, 1949), are of little use for shock absorption studies since they were obtained by static testing. Jorgensen et al. (1989b) developed a test consisting of a constant velocity compression to study heel pad properties. This method produced good correlation between the total deformation and force attenuation in a drop test over a force plate with cadaver samples and, as a consequence, Jorgensen used total deformation as a measure of shock absorbing ability of the heel pad. However, any test using a constant velocity that was slower than normal walking velocity would be inappropriate for testing viscoelastic materials and, furthermore, a cadaver heel pad may lose its properties due to fluid extrusion during testing.

Noe et al. (1993) used an impact hammer to test a cadaver leg suspended from wires with an accelerometer mounted into the tibia. Transmission of acceleration to the tibia after hammer impact was used to assess shock absorption.

Work based on dynamic methods using drop and pendulum devices have used peak deceleration, peak displacement, time-to-peak deceleration and percentage of input energy absorption, as common parameters, to study heel pad properties (Cavanagh et al., 1984a; Jorgensen et al., 1989b; Kinoshita et al., 1993a; b; Valiant, 1984). Kinoshita et al. (1993) also studied the time-to-peak displacement. Valiant (1984) carried out a more detailed description of heel pad properties under impact loading. He included the above parameters and others resulting from a more detailed analysis of the force - displacement response of the heel pad and showed an initial low stiffness region followed by a maximal stiffness. Displacement of heel pad continued after peak force had been reached which was related to the viscoelastic nature of the heel pad. Valiant computed initial stiffness, maximal stiffness, percentage of force at peak displacement, percentage of displacement after peak force and the time from peak to zero force level to describe these features (Figure 7.16). Stiffness at a force equal to body weight has been also studied in some running studies. However, during walking the forces do not reach body weight so this parameter cannot been calculated for walking (Aerts & De Clercq, 1993b).

It is reasonable to suppose that many of these parameters are correlated to some extent and probably describe different aspects of the heel pad nature under impact loading. Besides, a clear relationship between all parameters as well as with heel pad mechanics has not been established. On the other hand, some aspects of the heel pad mechanical behaviour have not been considered. Thus, more parameters could be defined in relation to heel pad behaviour. For example the duration of initial low stiffness displacement could be of relevance to initial loading for shock absorption. A medium stiffness region is observed after maximal stiffness and before peak force is reached which could be related to bottoming out of the heel pad that limits the contribution of the heel pad to shock absorption (Aerts & De Clercq, 1993b; Robbins et al., 1989). Nevertheless, the time delay between peak force and peak displacement appears to be a more consistent way of describing the viscoelastic nature of the heel pad. This is in accordance with the definition of the "loss tangent" parameter which describes the phase delay between displacement and force in the frequency domain (Forner et al., 1995; García et al., 1992, 1994).

The behaviour of the heel pad under impact loading [Chapter 2] depends on the heel's highly specialised anatomical structure and its mechanical integrity (Jahss et al., 1992a; Jorgensen et al., 1989b; Jorgensen, 1990). The main structural factors known to influence the mechanical behaviour of the heel pad are the size, location and integrity of the fat trabecular structure, the thickness of the fat pad, the amount of movement constraint and, possibly, the constituents of the fat itself [inferred from studies on degeneration due to ageing (Jorgesen, 1989b; Kinoshita, 1995)]. It has also been reported that insufficient heel pad function due to trauma or degeneration can lead to the development of shock-dependent injuries. In this sense, soft heel pads have been observed in a number of patients with heel pain (Khuns, 1949), plantar fasciitis (Sewell et al., 1980) and Achilles tendonitis (Jorgensen, 1985).

It is reasonable to suppose that individual variations in structural factors have an influence on heel pad properties. On the other hand, these factors are likely to show common patterns for population groups according to gender, age, race and others. Studies on heel pad behaviour under impact loading found in the literature show a great inter-individual variability (Jorgensen & Bojsen-Moeller, 1989a; Kinoshita et al., 1996a, b). There is evidence that demonstrates the dependence of the mechanical properties of the heel pad on several human-related factors such as age (Jorgensen, 1985; Khuns, 1949; Kinoshita et al., 1996a), gender (Jorgensen et al., 1989b), body fat content and others factors. However, the literature usually refer to tests with a small sample of participants focusing on the influence of isolated individual characteristics such as age or gender (Jorgensen et al., 1989b; Kinoshita et al., 1993a, b, 1996b), which makes the interpretation of results difficult and a comparison of studies virtually impossible.

Age related changes seem to affect the heel pad properties. It is believed that the adipose tissue of the heel pad is firm and pliable in the young and thinner and softer in the aged (Khuns, 1949; Perry, 1983). Prichasuck et al. (1994a, b) found that people aged 40 - 60 presented a thicker heel pad and a greater compressibility index (thickness loaded/thickness unloaded) (of 19.27 and 0.55 mm respectively) compared with people aged 20 - 35 (18.27 and 0.51 mm respectively) indicating a stiffness increase with age. These are "static" stiffness values and "dynamic" stiffness is expected to be greater since the viscoelastic properties of the heel pad usually show greater stiffness values under dynamic than under static loading.

Jorgensen et al. (1989b), in a constant velocity compression-decompression test (1.1 mms^{-1}, 1/1000 of the actual impact velocity), observed a reduction of total deformation with age, which was interpreted as a shock absorption decrease and a stiffness increase. However, a test at a constant velocity lower than actual is not suitable for measuring impact properties of frequency-dependent materials such as the heel pad.

Kinoshita et al. (1996a), in an *in-vitro* drop test, compared the heel pad properties of 20 elderly and 10 young people. The elderly people exhibited higher peak deceleration and smaller peak deformation compared with the young adults at a fast impact velocity (0.94 ms^{-1}). In addition, the ability to absorb energy reduces with ageing (78.1% for young and 73.6% for elderly). At a slow impact velocity (0.57 ms^{-1}), only lower energy absorption in young was observed (76.5% for young and 73.8% for elderly). These results indicated an age-related decline of the shock absorbency of the heel pad in response to a relatively fast impulsive input force. The heel pad in the elderly, therefore, appeared to maintain its function relatively well with only a slight difference from that of the young adults when attenuating low impact shocks comparable to those encountered during walking.

Differences have been also found between children and adults with children showing a less efficient heel pad (Kinoshita et al., 1993b). An *in-vitro* drop test showed higher peak forces, shorter time-to-peak, lower deformation, shorter time-to-peak deformation and lower energy absorption in children (73.9% in children vs. 78.8% in adults). However, it has to be remembered that results from drop tests reported in the literature tend to be lower than results from ballistic pendulum tests and the position of the participant was far from physiologically ideal, which could affect results.

Gender differences in heel pad properties have also been found. The heel pad is thicker in males (19.42 mm) than in females (18.12 mm). Jorgensen et al. (1989b) described a significantly higher total deformation of heel pad in men than in women. However, there is no information about gender influence in heel pad properties from dynamic tests.

The influence of obesity has been studied using skinfold measurement techniques to estimate body fat content (Kinoshita et al., 1993; Valiant, 1984) and the Body Mass Index (BMI is calculated by dividing the individual body weight in kilograms by the square of height in metres). Individuals with a BMI greater than 27 are usually considered to be overweight. Data on fall-related hip fractures on elderly women showed that the higher the BMI, the lower the forces in the hip due to a greater shock absorption at hip soft tissues level (Robinovitch et al., 1995). Kinoshita et al. (1993b) found a significant positive correlation ($r = 0.57$) between shock absorption and average skin-fold thickness measurements concluding that the thinner the body the less energy was absorbed. Prichasuck et al. (1994a, b) measured a greater compressibility index (i.e., greater stiffness) in heel pads of overweight humans.

It has also been proved that the **heel pad thickness** has an influence on heel pad properties. According to Jorgensen et al. (1989b) the thinner heel pads have lower shock absorption whereas the thicker heel pads show greater stiffness. A thicker heel pad has been observed in people suffering from achillodynia (Jorgensen, 1985), unilateral plantar heel pain (Amis et al., 1988) and acromegaly (Steinbach & Russell, 1964), whilst foot ulceration in diabetics has been related to the loss of heel pad thickness (Gooding et al., 1986). Thickness has been shown to depend on age, gender and body fat content (Prichasuck et al., 1994a, b). However, its influence has not been assessed using dynamic tests.

On the other hand, Valiant (1984), using an impacting pendulum, studied the difference of heel pad characteristics between runners and non-runners. No significant differences were found. Despite this, two different force-time and four force-displacement curve patterns were identified without any relationship with gender, age or per cent body fat content (measured by skin fold techniques), suggesting a more complex individual-dependence of heel pad properties.

It is evident, therefore, that there is great confusion about the dependence of the heel pad properties on individual characteristics; no clear relationship has been established (Jorgensen, 1985; 1990; Prichasuck et al., 1994a, b). Besides, most of literature focuses in running but little in walking. At the same time, little research has been devoted to study the relationship between heel pad mechanical parameters and impact forces. In consequence, a deep analysis of heel pad mechanics using a reliable testing method and experimental protocol was necessary before testing the influence of footwear in heel pad properties.

According to previous work (Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1996a, b) with *in-vivo* drop and pendulum tests, it has to be assumed that heel pad properties were not isolated from the contribution of the lower leg in absorbing impacts. This was appropriate given that the final interest of these investigations was really lower leg shock attenuation. Therefore for simplicity and consistency with previous work, the mechanical properties of the lower leg will be referred to as heel pad properties.

This Chapter describes work conducted to investigate heel pad properties from impact mechanics point of view to ascertain the role of the heel pad in shock absorption and the influence of gender, age and obesity. This was the first stage in exploring the statement that footwear influences natural shock absorption. A literature survey is first presented including a comparison between different testing methods.

7.2. Methods for the analysis of shock absorbing properties of the heel pad

A reliable test method was needed to facilitate analysis of heel pad mechanics. An extensive number of approaches and methods for measuring heel pad properties under impact loading appear in the literature. In a general sense, the methodology described in Chapter 3 could be used for heel pad analysis if differences arising from working with a biological system are taken into account.

Both human and machine testing methods have been used to study the heel pad. The main problem with the human methods is the lack of control of initial impact conditions, making difficult the understanding of the mechanics of landing (Lafortune & Lake, 1995a); it is also very difficult to isolate the contribution from the different shock absorbing mechanisms of the human body. As a result machine testing has been more widely used, drop and pendulum devices being the most employed methods. Machine testing can be used both *in-vitro* [live humans] and *in-vitro* [cadaver samples].

The simplest method used for heel pad characterisation is by palpation. This method is based on the fact that the mechanical integrity of the heel pad is related, to some extent, to the facility to manually compress it (Jorgensen, 1985). Whilst this may be of some utility in a clinical setting, more technical methods are needed to study impact mechanics. The different methods found in the literature report different results and the greatest discrepancies are between *in-vitro* and *in-vitro* methods. Despite this research, little attention has been paid to the behaviour of heel pad under load during walking. A review of the literature on these methods is presented below for the purpose of choosing the most suitable method to pursue this research. Methods found in the literature may be divided into **machine** and **human** procedures.

7.2.1. Methods for machine testing of heel pad

Machine testing of human heel pad includes *in-vitro* and *in-vitro* methods. In generally, provided the sample [human or cadaver] can be adequately positioned in the testing device any of the test methods described in this Chapter could be used *in-vitro* and in-vitro. However, results from *in-vitro* and in-vitro show significant differences (Aerts et al., 1993a; Aerts & De Clercq, 1993b, c) and these are discussed later.

7.2.1.1. Drop test

This method has been widely used for testing materials and biological systems such as knees (Hoshino & Wallace, 1987). In fact, standard methods for measuring shock absorption of surfaces and footwear materials are based on the drop test principle.

In drop tests, weight of known shape, dimensions and mass is allowed to fall from a controlled height onto the material under test. Instrumentation includes an accelerometer mounted on the falling weight (to register acceleration during impact) and displacement transducers or force plates. The input energy is equal to potential energy change and drop height and missile weight can be changed to configure the test for a desired impact velocity and input energy. However, impact force and the time of application are not controlled which makes it difficult to simulate real loads.

The drop test has been used for characterising the heel pad both *in-vivo* (Kinoshita et al., 1993a, b, 1996a, b) and *in-vitro* (Jorgesen et al., 1989b). Different test set ups and parameters of study are found in the literature.

Jorgensen et al. (1989b) carried out *in-vitro* tests on heel pads. A force plate located below the test sample was used for measuring impact force and time, allowing control of impact quality. The mechanical behaviour of the heel pad was described by the relative energy absorption as the ratio of impact force reduction and by the energy dissipated during impact which was calculated from the acceleration signal as the difference between the drop and rebound height after impact height.

Some insole materials such as EVA (Ethyl Vinyl Acetate) and Sorbothane were also tested using an impact velocity of 1.4 ms^{-1} (corresponding to fast walking), the contact area was 9 cm^2 and impacting mass was 1.6 kg (corresponding approximately to tibia and foot weight). The stiffness of the sample was calculated from the equation: *Impact force = Impact velocity x SQR (K_h x missile mass) where K_h is the stiffness.* Stiffness resulted in 8.2 x 10^5 Nm^{-1} for a sample of Sorbothane 1 cm thick, 2.4 x 10^5 Nm^{-1} for a sample of EVA 1 cm thick and 1.9 x 10^5 Nm^{-1} for a heel pad 1.4 cm thick. The energy absorption was 86 % for the heel pad, 85% for sorbothane and 60% for EVA. The heel pad presented a ratio of mean impact force reduction of 1.13 and of 2.09 for EVA and Sorbothane respectively. The same authors (1989a), tested six cadaver heel pads using a similar procedure. The transmitted peak force was registered as a measure of shock absorption with values ranged from 620 N to 900 N. According to Jorgensen greater differences should be expected in *in-vitro* tests, bearing in mind the viscoelasticy of the heel pad.

Kinoshita et al. (1991, 1993a, 1993b, 1995) carried out *in-vitro* studies of heel pad properties using the drop test method. The participant was positioned horizontally on stomach with the right knee flexed at 90 degrees to the horizontal. The ankle was plantar-flexed at 90 degree and fixed to a support rig with nylon straps (Figure 7.1).

The impacting mass was a steel rod 80 cm long weighing 5 kg. The impacting area was circular of diameter of 40 mm (12.57 cm^2). An accelerometer and a displacement linear transducer (DLT) were mounted on the impacting mass.

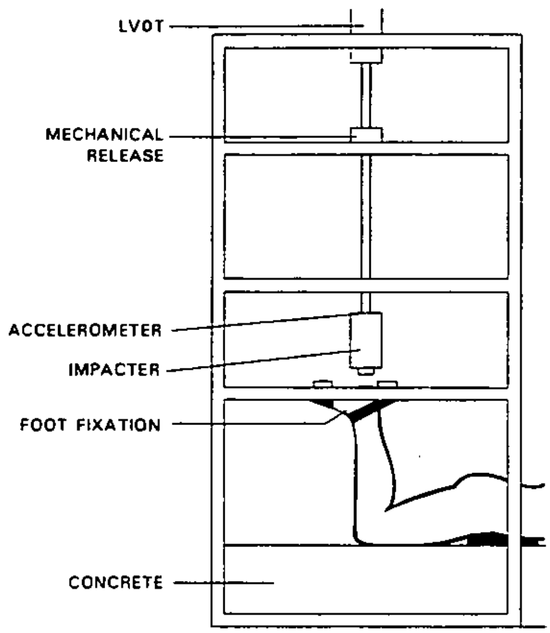


Figure 7.1. Drop test set up.

Fourteen adults and three children participated in a first experiment (Kinoshita et al., 1991). Impactors of 8.5 g with a 30 mm drop height and 11 g with a 50 mm drop height were applied for adults but only 8.5 g impactors with a 30 mm drop were performed on children. Forces in the range 416 N to 536 N were registered. Similar tests on samples of EVA resulted in impactor's acceleration of 11 g and 13 g for both drop heights respectively. Thus for the same input (drop) energy, lower impact forces (mass x acceleration) were registered for the heel pad, reflecting a greater shock absorption ability of the heel pad. The maximum displacements recorded were 10.1 mm and 11.7 mm respectively for each drop height.

The energy absorption was the area enclosed in the hysterisis loop observed in the force-displacement curve. For adults the average energy absorption values were 76% for the lower drop height and 77 % for the higher drop height (energy levels of 427N and 569N peak forces respectively). Lower energy absorption (74%) was reported for children.

Using the same procedure Kinoshita et al. (1993a, 1993b) conducted a more comprehensive study on the differences between adults and children. The effects on heel pad properties of repetitive loading and of running 30 km were also studied.

In the repetitive loading experiment (with ankle in slight dorsiflexion from 5 to 10 degrees), two different drop heights were used-for different impact velocities (0.95 ms^{-1} and 0.72 ms^{-1}) on nine females and seven males together and five seven-year-old children. Five impacts from each drop height were performed with each adult and five impacts from the lower height for the children. Three adults participated in the analysis of repetitive loading with an impact frequency of one every ten seconds for 6 minutes. Finally, two habitual runners were subjected to impact tests before and after running 30 km; the heel portion of a running shoe comprised three different layers namely one of EVA 25 mm thick, one of rubber of 7 mm and a third layer of EVA 9 mm thick.

High repeatability of the method was reported (0.86-0.93 test-retest method). Peak deformations of 9.5 mm and 11.3 mm for impact velocities of 0.72 ms^{-1} and 0.93 ms^{-1}

respective were measured. Peak force increased with velocity but time-to-peak decreased. From the load-displacement curve, the heel was observed to continue deforming even after peak deceleration was reached. On the other hand, a residual deformation of about 6 mm was observed at zero force level reflected the viscoelastic nature of the heel pad. Energy absorption showed a large inter-individual variability: values of 77.4% and 78.8% were measured for impact velocities of 0.72 ms^{-1} and 0.93 ms^{-1} respectively. Repetitive loading and running appeared to have no effect on heel pad properties. The shoe caused higher peak force and lower energy absorption (58%) than the heel pad. Similarly, child heel pads appeared to be less efficient than adult heel pads since lower energy absorption of 74% was measured

Kinoshita et al. (1995) used *in-vitro* drop test to study age related changes in the mechanical properties of the human heel pad. The same procedure as in previous studies was used, namely a 5 kg dropping mass with 40 mm diameter circular contact area. Impact velocities of 0.57 ms^{-1} and 0.94 ms^{-1} were performed on three age groups: 10 young adults, 10 active elderly aged 60-67 and 10 active elderly aged 71-86. The ankle was dorsiflexed at 5-10 degree from the neutral anatomical position. Ten impacts from each drop position were performed on each participant.

There were significant age effects for both peak force and peak displacement at the faster impact velocity. The elderly experienced higher peak force and smaller peak deformation (11.91 g and 10.36 mm for the 60 – 67 age group and 12.57 g and 10.4 mm for the 71 – 86 age group) than the younger participants (10.84 g and 11.2 mm). No age effect was found for peak force time and peak displacement time. Force-displacement curves shared common features for all age groups and these were in agreement with work of others (Cavanagh et al., 1984; Valiant, 1984): low initial stiffness followed by a high-maximal stiffness, deformation followed after maximum peak deceleration and residual deformation observed at zero force level.

There was a significant age effect on energy absorption at the two impact velocities. At the lower velocity impact (0.57 ms^{-1}), mean energy absorption was 73.8% (range 69.6 - 84.5%) for the group aged 60-67 years, 72.6% (65.4 - 78.4%) for the group aged 71-86 years and 76.5% for the youngest age group. Mean energy absorption values for the faster impacts were 73.65%, 72.6% and 78.1% respectively.

Whilst the drop test has the advantage of simplicity of construction, the position of participants is quite uncomfortable and the orientation of the heel pad is less physiologically acceptable than in others methods. Also, the plant is faced upwards and may alter the normal function of heel pad structure.

7.2.1.2. Human Drop

This is an *in-vitro* test where a person is dropped from a given height over an impact surface (Jorgensen & Bojsen-Moeller, 1989a) or simply hits a force plate from a sitting position (Nigg & Denoth, 1980). It is a constant energy method (equal to potential energy change). Knee and ankle are clamped to avoid shock absorption from joint mobility. Impact force is measured by means of a force plate and the acceleration transmitted to the person is registered using accelerometers located at different anatomical sites.

Jorgensen et al. (1989a) used this test for measuring shock absorption by comparing impact forces arising from different test conditions. Nigg & Denoth (1980) reproduced barefoot running forces and both ground reaction force and acceleration in the leg were registered. Displacements in the leg, calculated by the double integration of acceleration, were of 4mm and 10mm for forces equal to body weight and twice body weight respectively. Leg displacement was the result of ankle movement and deformation of soft tissues of heel pad and lower leg.

Denoth (1986) used this method to analyse heel pad mechanics in four population groups from six years to 80 years of age with the test participant sat on a chair and allowed to hit the force plate with the heel maintaining the foot in dorsi-flexed position. Reaction force, acceleration of the rigid part of the lower limb and the impact velocity were measured. Double integration of acceleration was used to obtain force-displacement curves, but soft tissue deformation and possible movements in the ankle joint contributed to final results in an uncontrolled manner. Energy absorption of about 90% and displacements between 4 mm and 10 mm for forces between body weight and twice body weight were obtained. Lower peak forces were observed in the elderly and in young children compared with other adults and with teenagers, whilst peak displacements showed no great differences.

This method has little control over impact conditions and since, the whole body is involved it is difficult to analyse results but this method could yield good results for measuring shock transmission.

7.2.1.3. Impact hammer

The impact hammer with accelerometer mounted is a device frequently used for modal analysis and for calculating transference functions of mechanical systems. A pulse of a pre-determined form, depending on the shape and material of the hammer's tip, is input to the system. Impact hammer has been widely used in biomechanics to measure bone properties [Hopkinson's stress bar technique] where the Young modulus of bone is calculated from the velocity of propagation of a stress wave input of the impact hammer.

Noe et al. (1993) used this method to study *in-vitro* properties of heel pad. An accelerometer was mounted on the tibia of a cadaver leg horizontally suspended from four wires to secure horizontal oscillation of the leg. The hammer was suspended also from a wire. The weight of the hammer was 0.45 kg with a circular impact surface 2.54 cm in diameter. The input energy was 0.83 J and the impact velocity 1.9 ms^{-1}. Impact of a steel bar was used for validation. The author pointed out the need of an *in-vitro* test.

This method may be regarded as a simplification of the ballistic pendulum, but limited control of impact and reduced weight makes it impossible to study the heel pad under real loads. Results from this method focused on shock attenuation; peak acceleration was attenuated 80% by the heel pad and 55% by a sample of EVA (Ethyl Vinyl Acetate).

7.2.1.4. Instron Universal Dynamic Testing Machine

The Instron universal dynamic testing machine is widely used for the dynamic testing of materials (García et al., 1992, 1994). Load and displacement can be controlled and registered and the simulation of different loading histories can be programmed and force-displacement curves can be obtained as a result. This method has been applied only *in-vitro* on isolated heel pad cadaver samples exposed to cyclic loading (Figure 7.2). This device allows accurate reproduction of walking loading history (García et al., 1994) and is best for testing non-linear and viscoelastic materials, however it would be very complex to adapt the Instron Machine for *in-vitro* testing.

Early work using this method was on the mechanical properties of the pads of different mammals (Alexander et al., 1986). Using the same method, Ker et al (1989) tested five human pads (mean thickness 16.3 mm) attached to the calcaneus. An Instron 8031 dynamic testing machine was used to cyclically apply a sinusoidal loading between zero and 2 kN (corresponding to running loads) at frequencies from 0.22 to 5.5 Hz to the test samples. The authors were only interested in the loading phase to validate a mechanical model. Energy needed to compress the pads was 0.75J, independent on calcaneus presence.

Later work (Bennet and Kerr, 1990) using the same procedure on human heel pads revealed lower energy absorption than that calculated in *in-vitro* tests, i.e. 30% compared with 80-90%. These differences were attributed to methodological errors and the presence of the lower leg *in-vitro* tests. Bennet and Kerr (1990) tested eleven samples applying 2 kN at frequencies ranging between 0.1 and 70 Hz. A non-linear stiffness of 1160 ± 170 kNm^{-1} was observed at a force equal to body weight. Energy dissipation ranged between 20.8% and 40.7%. No influence of loading rate on results was observed.

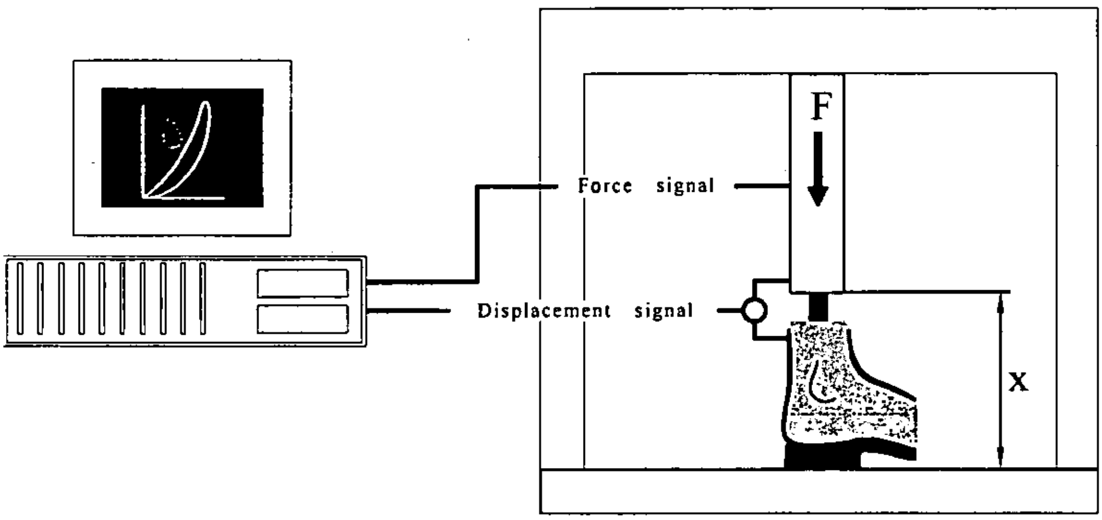


Figure 7.2. Scheme of e dynamic testing machine

Following some methodological improvements, a deeper comparison between *in-vitro* methods (pendulum) performed *in-vitro* and Instron tests was completed. Although differences were observed, results from the Instron were closer to pendulum *In-vitro* results (48% energy absorption vs. 50% respectively) (Aerts et al., 1995).

7.2.1.5. Heel Pad Compressor (HPC)

The HPC was developed by Jorgensen et al. (1989b) as a simple device for quantification of the heel pad shock absorbing properties in clinical setting and to investigate the significance of the heel pad thickness on these properties. This method included a device to prevent movement by fixing the foot and the shank at right angles to enable measurement of actual heel pad properties. A standardised impact was applied through an electromotor that drove a piston with a contact area of 9 cm^2, which corresponds to the minimal area of heel contact in heel strike running.

Cycles of compression-decompression loading were applied to the heel at constant velocity of 1.1 mms^{-1} (1/1000 of the actual impact velocity) as limited by the piston. The method was tested by comparison of results from tests in HPC and drop test on six cadaver human heel pads. The method was based on analysis of the force-displacement characteristics of the heel pad. Maximal deformation, compression energy (loading curve area) and the difference between area of loading and unloading curves as an expression of shock absorption were calculated.

Repeatability (1 - COV, where COV is coefficient of variation and equal to standard deviatioNm-1ean) and the effect of repetitive testing as well as the influence of gender, age and thickness were assessed by testing 100 average persons (50 men and 50 women, age range 17-84 years). X Rays were also used to measure thickness and weight bearing compression of the heel pad.

Poor repeatability of unloading data was apparent, so only compression results were used which would invalidate shock absorption studies using this method. However, a high correlation was found between peak force transmitted (loosely related to shock absorption) measured on drop test and HPC maximal deformation. Therefore the peak force transmitted parameter was selected for clinical use as an expression of shock absorption of the heel pad. Nevertheless, correlation was found in *in-vitro* test on cadaver samples that probably lost their viscoelastic properties.

This method uses a constant and slow velocity loading - unrealistic for viscoelastic materials such as the heel pad. Since the properties of viscoelastic materials vary significantly with displacement velocity. On the other hand, the origin of the relationship between total deformation and force transmission in the drop test and its significance for shock absorption measurement it is not clear (Jorgensen et al., 1989b). Thus the correlation between drop test and this method should be extrapolated with care due to the viscoelastic phenomena.

A mean total heel pad deformation of 6.7 mm was found (range 3.2 mm to 10.5 mm). Total deformation decreased with age and was significantly higher in men than in women. The mean body-weight load compression was 10mm.

7.2.1.6. Human pendulum

This procedure has been recently developed (Chavet et al., 1997; Lafortune et al., 1995b) as an alternative to other *in-vitro* machine testing procedures where real impact forces (generally for running conditions) cannot be simulated without causing pain. This involves a participant, lying on lightweight bedding dropped so that the right foot hit a rigidly wall-mounted force platform (Figure 7.3). A miniature accelerometer was mounted at the distal end of the shank for measuring shock transmission. This method allowed to some extent reproduction of forces encountered in *in-vitro* human locomotion in a controlled manner.

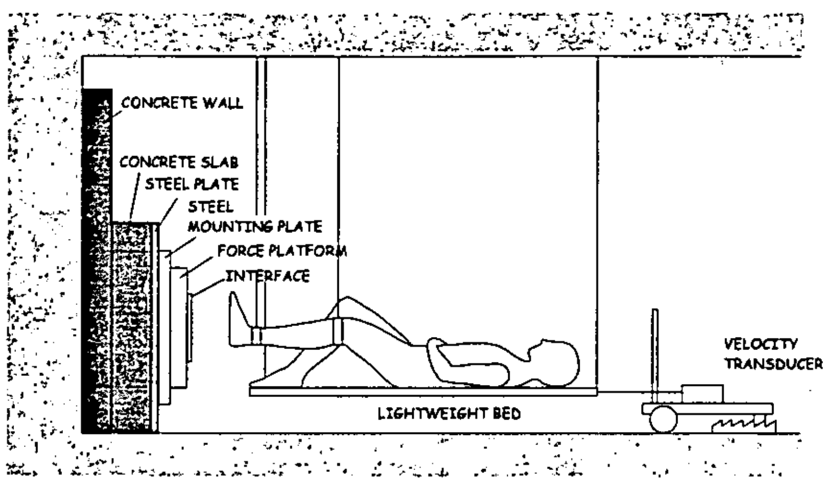


Figure 7.3. Human pendulum test

Ten healthy males participated in an experiment with an impact velocity of 1.15 ms^{-1} (running). With maximum knee extension and controlled ankle dorsiflexion (Lafortune & Lake, 1995a), peak value, time to peak and rate of loading were computed for both wall reaction force and shank acceleration. Fast Fourier Transform (FFT) was used to determine the power spectra of both signals for comparison with reported running values. Results were quite similar to those in the running experiments. A good simulation of reaction forces and tibia accelerations and a low Coefficient of Inter-individual Variation was shown (lower than 2%) for both.

In a more recent work (Chavet et al., 1997), this method was used to assess if human cushioning properties were similar for left and right legs. A bi-axial accelerometer was also used in this study to register head acceleration. Twenty-four humans took part in the experiment; time and frequency domain parameters were analysed showing a clear asymmetry for most of parameters characterising shock absorbing (force reduction) and attenuating (reduction of shock transmission) properties of human body. Initial kinematics influence in shock absorption was avoided by controlling initial impact conditions.

The main advantage of this method over the ballistic pendulum is the possibility of applying higher forces and that it could be useful for running or jumping studies, but a rather complicated testing rig is needed. On the other hand, the supine position of the participants makes it more difficult to understand the role of the different natural systems since the whole body is impacted and muscular action is not clear. This method, however, shows clear advantages for studying shock transmission through the body to the head.

7.2.1.7. Ballistic pendulum

This method has been widely used for the study of heel pad properties (Aerts et al., 1995; Cavanagh et al., 1984; Valiant, 1984), as well as for the study of footwear materials (Aerts & De clercq, 1993b), force attenuation in trochanteric soft tissues (Robinovitch et al., 1995), fracture energy of different bones under impact loading (Currey, 1968; Nikolic et al., 1975) and impact behaviour of different biological tissues (Thurlow, 1963).

This method was developed as early as in 1984 by Cavanagh et al (1984) to study heel pad mechanics, since then improvements and technical modifications have been introduced by different authors, some shortcomings of the method have been pointed out and solved, but it remains basically the same.

The pendulum has been applied for testing the heel pad both *in-vivo* and *in-vitro*, as well as barefoot and shod. Results from *in-vivo* tests differ from that obtained *in-vitro* and many researchers have developed their own ballistic pendulum, but little differences are observed in the testing procedure.

This method basically consists in impacting the test sample with an oscillating mass on which an accelerometer is mounted to register acceleration during impact. The test human is placed with the leg horizontal and the knee firmly pressed against a rigid wall (while comfortably standing on the other leg) to minimise movements which will result in errors that will be computed as heel pad deformations. Parameters of study are calculated from acceleration signal. Numerical integration is used for calculating velocity and displacement, although other instruments can also be used to directly measure both values.

Cavanagh et al. (1984) used a variable mass suspended from 1.22 m long nylon wires. The impact area was an aluminium circular plate 8.8 cm diameter. Photocells were also used to initiate data acquisition. The impact velocity was calculated from potential energy change assuming no frictional losses and impact at maximum horizontal velocity. Double integration of acceleration was used to calculate mass displacement. High speed filming was used for measuring leg displacements. Very small (2 – 3 mm.) displacements were recorded and considered in the analysis of heel pad displacement.

In this work a fixed impact velocity of 1.03 ms^{-1} was used, as registered from cinematic analysis of heel strike during running. For the selected velocity the runners' tolerance to force was reached for a mass of 4 kg, so the test mass ranged between 1.9 and 4 kg. As a result, the input energy ranged between 2.96 J and 4.26 J .

Results were presented for a mass of 1.9 kg at 1.03 ms^{-1} impact velocity. High acceleration peaks (20.8 g) were recorded. Lower forces than in running were reached, which would indicate that impact with a flat plane plate does not match real pressure distribution in the shoe, since in runners do not experience pain at heel strike. Energy absorption results were approximately 85%, which indicated that an effective shock absorbing system was present; and this would include heel pad, foot and leg. But separate contribution of each element could not be evaluated from the results of this study.

Average peak deformation was 1cm. This value, though calculated from acceleration signal, was later confirmed by measurement from double exposure strobe photography. Force-displacement curves showed an initial low stiffness region (0.19 10^5 Nm^{-1}), followed by a high stiffness area (1.30 x 10^5 Nm^{-1}); then, the heel pad continued deforming after peak force had been reached; after unloading, a residual deformation about 5 mm was observed at zero force level.

Stiffness estimates were 5 and 10 times lower than those quoted by Denoth & Nigg (1981). An increased velocity gives rise to higher peak force, higher peak displacement and greater energy absorption (5% higher). The rising phase of the force-displacement curve, however, remained unaffected by velocity change. In the same work, different mass-velocity combinations were used to produce the same input energy and minor changes only were observed in heel pad mechanics.

The effect of cyclic impulsive loading was also analysed. Twenty-five repetitive impacts performed at three seconds intervals presented no significant differences in results. High speed lateral films registered a maximum displacement of the leg of 3 mm at 1.24 ms^{-1} impact velocity; this 3 mm value was subtracted from the heel pad displacement calculated from acceleration, but force-displacement curves remained unchanged since leg movement was observed to occur later in the time frame.

Valiant (1984), in a thesis paper used a similar ballistic pendulum and testing procedure to study differences in heel pad properties between runners and non-runners. The effect of firm heel pad restraining was also analysed but no differences were observed. A possible explanation lies in the influence of restraining mechanism due to blood flow that seemed to negatively affect heel pad shock absorbing properties. This result is in conflict with those from walking studies (Jorgensen et al., 1988, 1989a) where heel pad function improvement was observed by heel pad confinement in shoes. However, in this work a firm compression was laterally applied to the heel which actually does not mimic heel confinement by a shoe heel counter.

A pendulum mass of 1.92 kg was used to test 24 participants. Double exposure stroboscopic photography reflective markers on the heel enabled three-dimensional movement of the heel pad during impact to be studied. The effect of velocity was analysed comparing results from tests at impact velocity from 0.8 ms^{-1} (fast walking) to 1.2 ms^{-1} (running). The results showed a loading velocity dependence - peak force increased from 223 N to 437 N as impact velocity increased. The energy absorption was calculated as the area of the hysteresis loop from the force-displacement, results ranged between 84% and 99% absorption, decreasing only 1% due to heel restraint. Maximum displacement of 9.9 mm was obtained, decreasing 1mm with heel restraint. Estimates for stiffness averaged 7.9 kNm^{-1} and 105.6 kNm^{-1} for initial and maximal stiffness regions, respectively.

As a conclusion, the non-linear and viscoelastic nature of the heel pad was stated showing a high shock absorption ability. The analysis of the force-displacement curve revealed a low initial stiffness, related to initial reshaping of heel pad in medial-lateral and posterior directions as reflected by X Ray results. Average displacements of 2 mm and 2.1 mm were observed in lateral and medial directions, respectively. This medial-lateral displacement occurred during the first 4 or 5 mm of forward movement of the pendulum against the heel pad. This reshaping was interpreted by Valiant as a mechanism of energy absorption. The author found no differences between the heel pad of runners and non-runners. However, different patterns of force-displacement curves were observed with no apparent relationship to gender or age. This suggested further and complex influence of individual's characteristics.

Aerts et al. (1993a, 1995) carried out pendulum tests with an impacting mass of 11.615 kg at an initial velocity of 1 ms^{-1} simulating running conditions. The results showed an energy absorption from 80% to 90% (De clercq et al., 1993; Aerts et al., 1995). **Aerts & De Clercq** (1993b) carried out also *in-vitro* pendulum tests on isolated heel pads to compare the results with Instron *in-vitro* tests. This work reinforced the proposition that *in-vitro* pendulum results are influenced by the contribution from other elements of the lower leg. This work also showed that pendulum test results were affected by the mass and rigidity of the supporting wall. A low rigidity wall could gain kinematic energy in the test, which could be interpreted as energy dissipation thus magnifying results. So, the support wall should deserve special consideration in the set up procedure.

Aerts & De Clercq (1993b) studied the behaviour of the heel pad in shod condition. According to the authors, small differences are found in ground reaction forces during running with soles of different hardness. Traditionally this has been explained by changes in initial pronation of the foot. However, the authors analysed changes in the heel pad behaviour in shod conditions as an alternative explanation. Drop and pendulum test on two shoes differing only in their midsole hardness revealed that the harder sole was five times stiffer than the other.

The pendulum test was used for analysing the shod condition. Ankle pronation was avoided by clamping the metatarsal-phalangical joint to exclude its possible effect on impact forces reduction (Denoth, 1986). The participant was asked to relax the lower leg muscles to avoid muscular tension on the foot. At the same time, the femoral condyles were firmly pressed against the wall by the hip flexors. In this way, shock absorption due to the lower leg was reduced.

The pendulum was suspended from 1.51 m from the foot using non-extensible cables, an impacting mass of 11.615 kg and a contact surface of 32.2 cm^2. Impact velocity for double integration was obtained from energy change calculations, low walking (0.37 ms^{-1}) and moderate running (1.06 ms^{-1}) corresponding impact velocities were extreme values and a 0.96 ms^{-1} impact velocity was finally used. Double integration of acceleration was used to obtain displacement. In that work, a linear steel spring was impacted to detect eventual resonance effects. A small vibration of a frequency 28 times higher than the impact frequency itself was observed. The calculated stiffness for the spring differed less than 2.5% from the calibrated value. The coefficient of variation for impact force obtained from 8 trials on a participant was only 2% which indicated the low variability of the method.

Nine male recreational runners participated in the experiment (22.3 years and 72.8 kg on average). All participants performed three tests at 0.96 ms^{-1} impact velocity for each shoe, whilst only one participant performed tests at all velocities between 0.37 ms^{-1} and 1.01 ms^{-1} (to assess velocity gradient effect). Load-displacement curve was also obtained for the isolated shoes using the pendulum to study heel pad shod compression. The compression of the heel pad was obtained by subtracting the actual deformation of the isolated shoe from the shoe-foot system deformation. The shoes were studied under the same loading conditions while supported by means of an artificial rigid lower leg. Sole thickness was about 20mm for both shoes. Results showed a stiffness of 67 kNm^{-1} for the soft sole and 314 kNm^{-1} for the hard sole (small differences were observed at different impact velocity). Additional experiments on a participant with a piece of EVA 19.6 mm thick glued on the heel pad. A stiffness of 130 kNm^{-1} was obtained for the isolated Eva sole.

Regarding the shoe-foot system, results in the 0.96 ms^{-1} experiment were significantly different in all the test parameters for soft and hard shoe-foot systems. The soft system presented a 40% lower peak load (614 N v 864 N), higher time-to-peak (22.4 ms v 15.9 ms) and higher maximal compression (15.2 mm v 11.8 mm) as well as a lower loading rate than the hard system. This result would support the theory of ankle mobility as a mechanism of midsole stiffness compensation. However, these differences were not shown for the 0.37 ms^{-1} experiments and this might suggest that for low loading the heel pad could act as a compensating mechanism. The authors found unexpected results when studying the heel pad compression on both: for equal energy input the heel pad always deformed less in the hard (6.7 ± 0.9 mm) than in the soft (7.6 ± 0.9 mm) shoe-foot system. According to mechanical concepts on springs arranged in series with different stiffness, stiffer springs show lower displacement thus a greater heel pad deformation in hard soles should be expected. Anyway, considering the relative contribution of heel pad to global compression it was more or less the same in both cases i.e. 6.7 mm out of 11.8 mm (ratio = 0.57) for the hard system and 7.6 mm out of 15.2 mm (0.5 ratio) for the soft.

In this sense, changing the impact velocity did not change the loading phase of the force-displacement curve and differences between hard and soft systems decreased as input energy (i.e. impact velocity) decreased, in such a way that important differences (40%) are found for running whilst no differences are found for walking. Differences in loading rate followed the same pattern. This would suggest a change in the behaviour of the heel pad: the lower the input velocity the better the heel pad would compensate for changing midsole hardness, probably related to changes in loading rate due to its viscoelastic nature. In barefoot studies, the maximal stiffness during loading increases with velocity. A shod heel pad did not present an initially low stiffness behaviour as described by others (Cavanagh et al., 1984; Valiant, 1984), probably due to some restraining effect of tight lacing of the shoes on the feet. In this sense, the initial stiffness was 100 kNm^{-1} for all the curves obtained for shoe-foot systems, similar to the results obtained for the isolated shoes. This would indicate that, at the beginning, only the shoe soles were compressed.

It was stated that lower leg compliance and its influence on results would be small given the viscoelastic nature of tissues under impact loading and for other reasons. At the same time, if present it should not affect heel pad deformation in shod condition. If lower leg compliance is present it should reinforce differences between hard and soft-shoes since hard shoe-foot systems are exposed to higher impacts and would be more affected due to higher forces and less viscous damping. This inverse coupling between heel pad compression and midsole hardness could be explained by considering the loading rate-dependent behaviour of the tissues of the heel pad. They stiffen as loading rate increases (viscoelastic behaviour); a higher loading rate was observed in the hard shoe causing a stiffer heel pad and lower deformation. On the other hand, results suggested that in the shoe the heel pad behaves more stiffly and showed a steeper loading rate dependency than barefoot. Consequently, a factor other than viscoelasticity must be also present, probably the effect of the design of the heel counter; a precisely fitting heel counter will restrict the heel pad counteracting the sideways expansion of the heel pad resulting in an increased stiffness and loading rate dependency. Results from testing the isolated piece of EVA (medium stiffness, no heel counter) presented larger maximal heel pad deformations than in any shod condition approaching barefoot results. That strongly reinforces the theory of negative effect of the heel counter on the heel pad properties.

Results strongly suggested that overall shock reduction in the hard shoe-foot system is lower than could be expected considering basic mechanical principles and the elastic behaviour of the isolated shoe. This is due to an increase in midsole hardness is accompanied by an additional stiffening of the heel pad due to its confinement into the shoe. This additional stiffening of the heel pad influences shock reduction negatively. These results contrast with Jorgensen et al (1988, 1989a) findings that showed an improvement of the shock absorbing ability of the heel pad by means of heel pad confinement. Differences may be explained in terms of mechanical overload due to heel pad bottoming out in drop test and in shod running tests of non-confined heel pads (used by Jorgensen) in such a way that later confinement avoided overloading thus improving so heel pad properties.

The conclusions of this work are (i) the heel pad does not stand up to the unexpected force reduction that takes place at heel strike when running with hard-sole shoes, favouring augmented ankle pronation theory, and (ii) a variety of factors such as maximal loading, rate of loading and heel counter effect determine shock absorption of shoe-foot system.

Aerts et al (1995) pointed out the unpredictable nature of pendulum since it is very difficult to obtain a perfectly centred, planar impact without pitching and yawing of the mass during the rebound. This causes energy loss that could explain some differences in results. The pendulum test is rather sensitive to the experimental circumstances such as wall effect, pendulum movement and drop position. However, it has been claimed that visual inspection is sufficient for securing the quality of the impact. On the other hand, spring and different mechanical tests have shown good repeatability and reliability of the method.

7.2.2. Methods for human testing of the heel pad

Movement analysis techniques have been widely used for studying shock absorption during walking and running (Forner et al., 1995; Jorgensen et al., 1988, 1989; Voloshin & Wosk, 1981, 1982; Wang et al., 1994). Some studied the reduction of impact forces and others the attenuation of shock wave, mainly regarding the effect of footwear and insole design and materials (Lafortune & Hennig, 1992; Light et al., 1980; Milgrom et al., 1985, 1992; Nielsen, 1990; Pratt et al., 1990; Rooser et al., 1988; Voloshin & Wosk, 1981, 1982; Wosk & Voloshin, 1981, 1985). Some work focussed in the effect of heel pad confinement on shock absorption and transmission (Jorgensen et al., 1988, 1989, 1990, Wang et al., 1994). More details on these techniques will be described later as part of the experimental work carried out in the course of this research.

In these methods the global effect of the wearer and footwear, as well as possible interactions within them are obtained. As a result, given the difficulty of isolating the heel pad from the others systems, few references to the heel pad properties using these techniques were found. Nevertheless, some interesting studies have been done.

Wang et al. (1994) used techniques for plantar pressure measurement to evaluate the cushioning effect of the heel cups, rubber or plastic housing for the heel. Increased shock cushioning related to the effect of heel confinement and to materials' properties were observed. The effect on rearfoot stability was also reported.

Jorgensen et al. (1988), using force plates measurement, reported a mean increase of 8.8% in energy absorption due to heel pad confinement, being significantly greater in people with reduced heel pad shock absorbency and for low impact conditions. Using accelerometers and electromyography, the heel counter effect was showed also to reduce heel strike transients, muscle activity and energy uptake (Jorgensen, 1990).

Several authors (Light et al., 1980; Voloshin & Wosk, 1981, 1982; Wosk & Voloshin, 1981, 1985) have used accelerometers to analyse differences in shock attenuation. However, few references, were found measuring deformation of heel pad and sole at the same time. Only De Clercq et al (1990) developed a technique to determine *in-vitro* deformation of the heel pad separately for midsole deformation and during a real running heel strike. A high speed (150 frames per second) cineradiographic filming was used in synchronisation with force plates. The movement was laterally filmed at 200 frames per second.

After film digitisation maximal deformation was measured to be (for two runners) greater in the midsole than in the heel pad. A very fast initial deformation of the heel pad was observed, remaining deformed even when the midsole was already recovering to its original shape. Thus, results showed the midsole to be more elastic than the heel pad (confirmed by the residual deformation shown by others). In any case, an increasing contact area for the midsole should be considered as the shoe gradually contacts the ground. The force-deformation curve was similar to that showed by previous methods with an initial stiffness of 0.51×10^5 Nm^{-1} to 2.31×10^5 Nm^{-1} and a final stiffness from 3.02×10^5 Nm^{-1} to 7.82×10^5 Nm^{-1} calculated.

In later work, De Clercq et al. (1993), analysed running barefoot and with sport footwear using force plates and registering deformation of heel pad and sole in two male runners using high speed video cameras and X ray film apparatus. This technique enabled the simultaneous registering of deformation of both heel pad and sole. The heel pad compression barefoot was 60.5 ± 4.4%, whilst shod was 35.5 ± 2.6%. This great difference was related to heel pad overloading barefoot.

The behaviour of heel pad shod was non-linear. Mean initial stiffness was (150 ± 11) x 10^3 Nm^{-1} and final stiffness was of (662 ± 255) x 10^3 Nm^{-1}. These values match only Denoth's results (1981, 1986). Cavanagh et al. (1984) registered however, a 5 fold lower stiffness with pendulum tests. Considering the heel pad and the sole as two linear springs in series, the contribution of the heel pad to global shock absorption will be through reduction of the system stiffness.

7.2.3. Comparison between methods

To choose a method for the analysis of shock absorbing properties of the heel pad, two major methodological aspects must be considered (1) machine or human testing, and (2) *in-vivo* or *in-vitro* testing. Shock absorption during shod walking is the result of the complex combination of natural (both active and passive) and external (artificial) systems, as well as the likely interaction between them.

Human tests present serious difficulties in isolating the heel pad contribution from the rest of systems (i.e. lower leg). **Thus, machine testing is preferred to study the shock absorbing properties of passive, both natural and external systems in general, and the heel pad in particular**. Human testing, however, remains useful for the analysis of active mechanisms as well as the global performance of systems combination.

Regarding machine testing, *in-vitro* and *in-vitro* tests are possible but substantial differences are found between them, although common features are observed in force-displacement curves obtained by both. *In-vitro* impact (drop or pendulum) tests on human heel pad show a large percentage of energy dissipation (76% to 95%) and a stiffness value of 150 kNm^{-1} for a load equal to body weight in the loading phase.

Results depend on loading rate. *In-vitro* tests, on the contrary, show energy loss of 30%, almost 10-fold stiffness (1160 kNm^{-1}) at body weight load and no relation between frequency of loading and mechanical behaviour of the heel pad (Aerts et al, 1993a, b, c, 1995; Alexander et al, 1986; Bennet & Kerr, 1990).

These differences are attributed to the mechanical behaviour of the entire heel-foot-leg-knee system involved in the *in-vitro* tests, which is called **"lower leg hypothesis"**. However fluid extrusion from the heel pad occurs during compression which may alter the mechanical properties of cadaver samples. For Bennet & Ker (1990) this possibility is very slight.

To analyse this hypothesis (Aerts et al, 1995) pendulum and Instron *in-vitro* tests were performed on the same isolated heel pad samples. Instron testing was modified from previous studies, where load-displacement curves were obtained after many sinusoidal cycles so that strain rate was maximal halfway deformation, completely different from pendulum test where maximal velocity was at initial contact and only one single loop as performed. So Instron tests were also recorded from the first cycle on, and with the actuator of the Instron hitting the sample half way along its sinusoidal displacement (i.e. at its highest velocity). Thus, Instron test could be better than the pendulum test since a realistic loading history can be applied.

Full cycle Instron test produced results consistent with previous Instron tests with energy loss of 30.9%, compared to near 80% from *in-vitro* tests. Half cycles test, however, showed a large variability in load-deformation curves, a dependence on Instron loading frequency and a higher energy loss, 48.2%, which reduced after the second cycle due to conditioning, thus confirming changes in the mechanical integrity of the test sample.

Although a substantial difference in energy loss persisted when compared with pendulum results, both methods revealed the same mechanics for isolated the heel pad to support the theory of the lower leg being involved in *in-vivo* measurements (Aerts et al, 1995). So, the lower leg hypothesis would be confirmed accounting for differences (between 25-40%) in energy loss. But, the greater differences observed by other workers may be due to differences in the loading regime which is very important given the viscoelastic nature of the heel pad. In this case, quicker and higher frequency impacts were performed with the pendulum (impact velocity of 0.51 m/s and time of maximum displacement (tdmax) of 13 ms than with the Instron testing machine (impact velocity of 0.37 m/s and tdmax of 23 ms). So it is reasonable to postulate that - according to previous chapters - if pendulum loading conditions were fully reproduced in the Instron, differences between both methods would be further reduced and a much lower leg effect would be observed.

On the other hand, *in-vitro* Instron tests allow fully controlled loading conditions - the main problem is to assure mechanical integrity of testing samples since changes due to fluid extrusion (conditioning) after the first test are observed. This should be carefully considered since the heel pad comprises soft tissues, with blood flow, and neural components responsible of mechanical response so *in-vitro* tests are unlikely to register actual properties. This makes it impossible to compare the results from machine and human testing which would be a way of gaining deeper insight on human shock absorbing mechanics. At the same time, working on cadaveric samples is an expensive and complex procedure.

So, from above analysis, *in-vivo* methods would be preferred. However, the literature survey presents three main shortcomings that have to be born in mind:

i. It is difficult to fully reproduce real loading since little control of impact is possible given that tests are mainly performed at constant potential (drop) energy.
ii. Peak force to be applied is limited by human's pain threshold making it impossible to reproduce running loads.
iii. The contribution of other natural shock absorbing systems such as leg, foot joints, etc. cannot be separated from heel pad ('lower leg hypothesis') (Aerts et al., 1995; Bennett & Ker, 1990) so shock absorption due to heel pad properties would tend to be over-estimated.

Pain tolerance will not be reached for walking loads and nowadays it is widely accepted (Aerts et al., 1995; Kinoshita et al., 1996a, b) that for the study of shock absorption during human walking the interest is on shock absorption of the lower leg without isolating heel pad properties.

So, for a large scale study on shock absorption during human walking on different populations an ***in-vivo* machine method would be preferred**. Regarding *in-vivo* procedures, quite a great variety of testing set ups have been used. Different loads, loading rates and testing methods, as well as means for controlling the contribution from joint mobility, muscular tension and other shock absorption systems are found in the different methods.

Given the viscoelastic (loading rate dependent) and non-linear (load level dependent) nature of the heel pad, the best would be to develop an Instron-like method to simulate real loading by load control which would be very expensive (testing people on the Instron is not possible). Considering the current methods, human drop shows control complications and human pendulum requires complex testing equipment and facilities whereas not many improvements are introduced for experiments under walking conditions, so either drop or pendulum tests would be preferred.

Drop and pendulum test results found in the literature differ by 10% for energy absorption (Bennet & Ker, 1990; Cavanagh et al., 1984b; Kinoshita et al., 1993a, b), probably due to methodological differences. A larger contact area can give rise to higher energy dissipation since a larger amount of fat tissue is compressed. Kinoshita et al. (1993a, b) found also that some participants showed a secondary peak on acceleration signal, which was probably due to ankle contribution because of free movement. Mass and velocity differ within methods although some references showed that different mass and velocity combinations which produce the same input energy resulted in similar energy loss (Cavanagh et al., 1984b; Kinoshita et al., 1993a, b;). Another factor contributing to the discrepancy may be racial differences in the thickness of the heel pad (Kinoshita et al., 1993a, b). It is known that the heel pad thickness affects shock absorption. On the other hand, the average body weight and height as well as the fat percentage in the Japanese population are about 5% less than those in the West, and this could explain differences between results from Kinoshita (Japan) and Cavanagh (Europe) and Valiant, Aerts *et al* (USA).

At the same time, in drop tests (Kinoshita et al., 1993a, b; 1995a, b) a mass impacts the heel pad facing upwards. This position seems not physiological since, due to gravity, 'pre-loading' soft tissues will tend to fall sideways on rest position affecting initial stiffness and reshaping and may influence final results (Valiant, 1984). Besides, the participant's position in the test rig is not comfortable and the regulation of people of different size is complicated. So, an instrumented ballistic pendulum was developed course of this research for *in-vitro* testing of the heel pad to study the passive shock absorbing properties of the lower leg under walking impact loading. This particular method was chosen because the position of the participant's leg in the test rig is more physiological than in other methods and a more comfortable position for the participant. Further, this method is simpler and cheaper than others and allows a design for comfortable positioning of people of different size.

7.3. Development of a ballistic pendulum for heel pad testing

This method consists in impacting the test sample by means of a free oscillating mass dropped from a certain height. It is a method of constant input energy, in such a way that all samples are tested under the same input energy (mgh), which will depend on impacting mass (m) and drop height (h). So, the weight of the impacting mass and the drop height can be fixed to obtain a desired impact energy, which allows certain flexibility to simulate real loading conditions.

The testing method developed in this work is similar to the used in previous works (Aerts et al., 1995; Cavanagh et al., 1984b; Valiant, 1984,), but some improvements have been introduced to overcome previous shortcomings. These shortcomings are due to earlier work being focused mainly on running conditions by adjusting drop height so to obtain an impact velocity similar to the vertical landing velocity of the foot. Few studies have focused in the behaviour of heel pad in walking.

Important aspects considered in the design and development of the method and testing procedure were:

i. *Contribution of other systems of the lower leg.* Active systems such as ankle pronation can be eliminated by blocking the joints at a given position. Regarding passive systems for the study of shock absorption during human walking, [agreeing with Kinoshita et al. (1996b)] the interest is on the shock absorption of the lower leg instead of only in heel pad properties; however for convenience and consistency with previous works, reference to heel pad (the main system) properties will be used.

ii. *Reproduction of force and duration of impacts.* Viscoelastic and non-linear materials require accurate simulation of loading history to study their behaviour in a testing machine. The pendulum does not allow this, but it permits quite a good reproduction of peak force and duration. This is a method of 'Constant Energy', that is the energy input is always the same according to the release position which actually happens when walking since the leg hits the ground falling from a given height.

iii. *Assure a physiological human position.* The participant's position in the pendulum was comfortable and quite similar to physiological position. Regulation was allowed.

iv. *The contribution of the supporting wall.* As described by Bennett & Ker (1990), a rigid supporting wall is required to avoid energy loss.

v. *Reliability and repeatability.* Pendulum results can be affected by errors due to differences in the drop height, non-planar impact and distortion of pendulum movement after impact. A good design of the pendulum support as well as visual inspection of each impact and resulting acceleration will minimise these errors.

vi. Assessment of *individual's characteristics influence.* Experiments were designed to assess influence of individual's characteristics on heel pad properties.

The literature review identified several problems with the pendulum method, which have been overcome in the method developed in this work.

i. Peak force, which can be applied, is limited by human's tolerance. This did not allow simulation of running heel strike, but the present study focused on walking conditions. Walking impact forces do not reach human's tolerance. Thus impact conditions could be simulated without limitations.
ii. Properties measured correspond to the lower leg and not only to the heel pad. In this work, it was assumed that properties measured correspond to the lower leg. Moreover, for the analysis of shock absorption during human walking the main interest is the analysis of passive properties of the lower leg and not only those of the heel pad.
iii. Pendulum is very sensitive to testing circumstances such as rigidity of supporting wall, errors in drop height and difficulties to perform an impact centred in the heel pad. A heavy stiff thick wall was used in this method. Besides, an important innovation of this method was possibility of regulation. The pendulum was designed to be easy and quickly fitted for people of very different size for improved impact quality.
iv. Velocity estimation by double integration from energy calculations can produce errors in heel pad properties. In this method, the impact velocity was indirectly measured instead of calculated from energy changes.

Important improvements of the method developed for this work were the parameters defined for describing heel pad properties, as well as statistical techniques used for the study of heel pad properties, which will be explained later. Methodological improvements that overcame the main errors observed in the literature, and others, will be also described later.

7.4. Design requirements

The main points considered in the development of this particular method were as follows:

1. *The participant:*

   Regulation: The method was to be applied for testing people of quite different size, from children to adults and elderty people.

   Comfort. Testing position had to be comfortable.

2. *Impact quality:*

   Assure that impact occurred at maximal horizontal velocity of the oscillating mass.

   Movement of the pendulum has to take place in a vertical plane.

Assure central impact to the heel pad.

3. *Reliability and repeatability of results:*

   Assure physiological position of the leg.

   Minimise frictional losses.

   Assure minimum contribution of the supporting wall.

   Control and avoid shock absorption due to the movement of knee, ankle and foot joints.

   Measure impact velocity.

4. *Test procedure:*

   Develop a quick and simple testing procedure.

   Assure reliability and repeatability.

   Possibility of using the method for testing materials.

   Possibility of changing the oscillating mass and drop height.

   Assess individual's characteristics to control their influence in results.

7.5. Description

The principle of this method consists of applying an impact to the participant by means of linear momentum change of an oscillating mass. The participant was comfortably placed in a structure with the leg horizontal and the knee firmly pressed against a rigid wall. An accelerometer was mounted in the impacting mass to register acceleration during impact (Figure 7.4). Impact forces were directly obtained from the acceleration signal whilst double integration was used for computing velocity and displacement.

The testing method developed in this work included both hardware and software. Hereafter, the hardware will be referred to as the pendulum. The pendulum comprises a structural part and associated instrumentation.

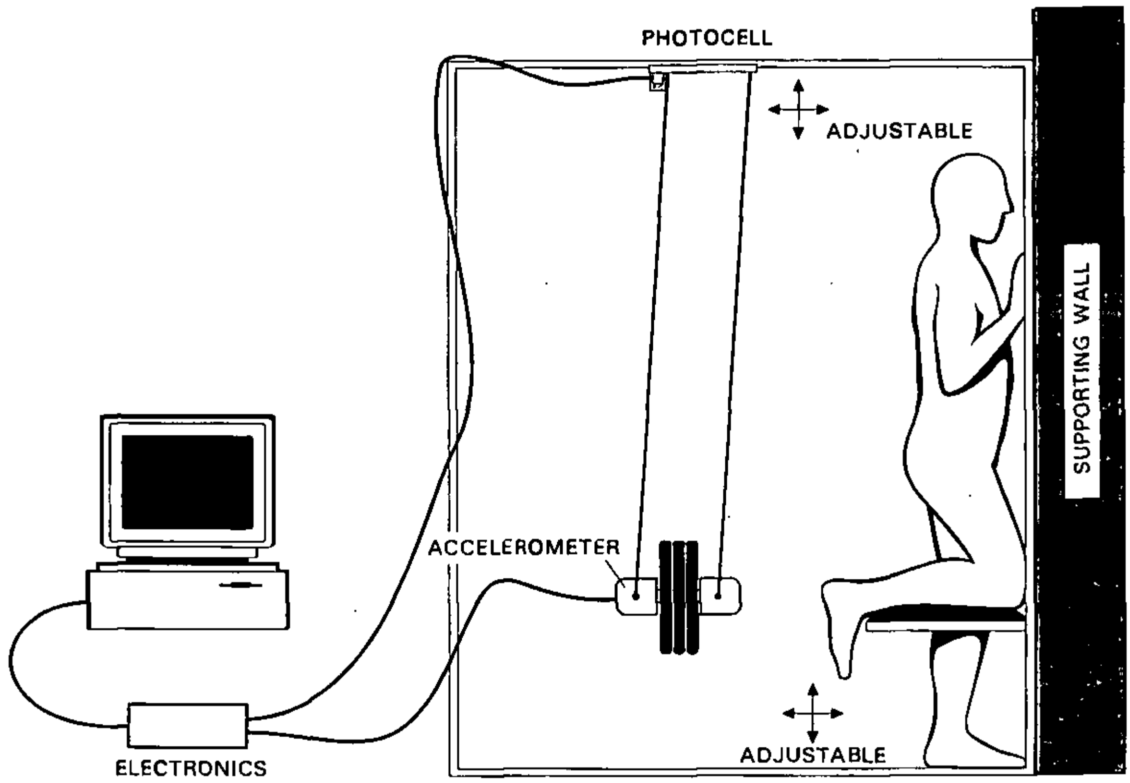


Figure 7.4. General testing set up

7.5.1. Structural part

The structural part incorporates the **test rig** and the **impacting pendulum.**

7.5.1.1. Test rig

The test rig was designed to provide support to the rest of elements and to accommodate the participant (Figure 7.5). It is a structure made from commercial aluminium bars. It consists of a prismatic frame of square base of side 120 cm and height 214 cm with a bar in the middle of one of the vertical sides used to mount the support for the participants and another bar was located at the top of the rig for mounting the pendulum.

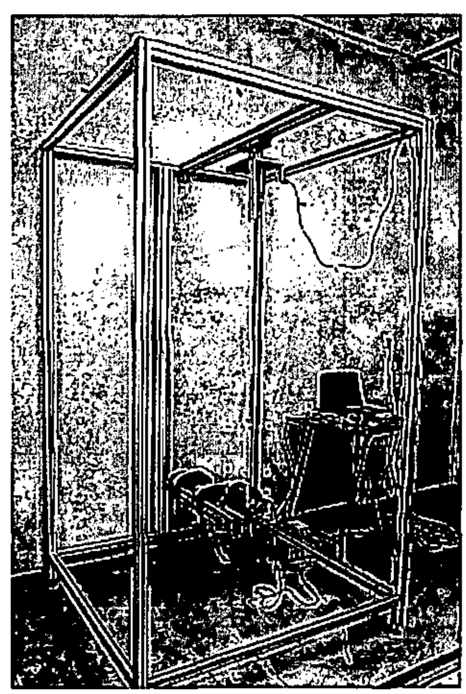


Figure 7.5. Test rig

The aluminium bars were of special profile with slots, which allowed easy mounting of instruments as well as low friction sliding of mobile parts. Dimensions and range of movement of the mobile parts were chosen to allow accommodation for people of different size (a detailed description is included in appendix A1).

The test rig was designed to be located against a wall (supporting wall hereafter) allowing the participant to rest at the testing position with no pain and damage in the knee. Two main structures were included in the test rig, one for the **participant's support** and the other for the **pendulum fixation**.

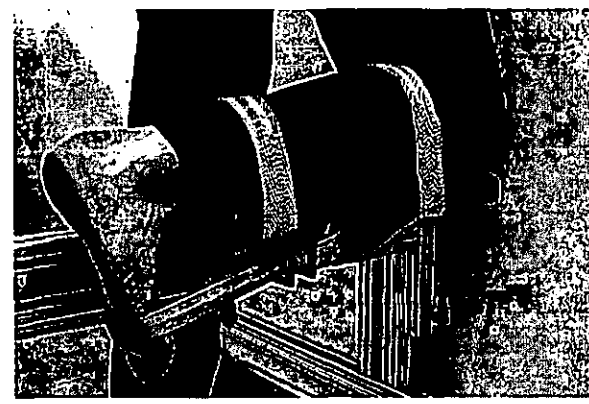


Figura 7.6. Leg support with detail of leg and foot clamping

The structure for locating the leg was designed to adjust to both participant's height and leg length. It consists of a horizontal bar adjacent to the central bar in the side of the main structure placed against the supporting wall. Two independent pieces of upholstered wood were mounted in this bar to cushion the participant's leg (Figure 7.6). These positions of these pieces along the horizontal bar are adjustable to provide good support for different size people. Two special screws secure them.

A hand support bar allows participants to comfortably support themselves against the wall. It is height adjustable along the central bar for a comfortable one-leg standing position during the test. The smaller piece (for ankle support) was thicker to compensate the irregular shape of the tibia and thus position it in a horizontal position and to ensure an impact perpendicular to the heel (Figure 7.7).

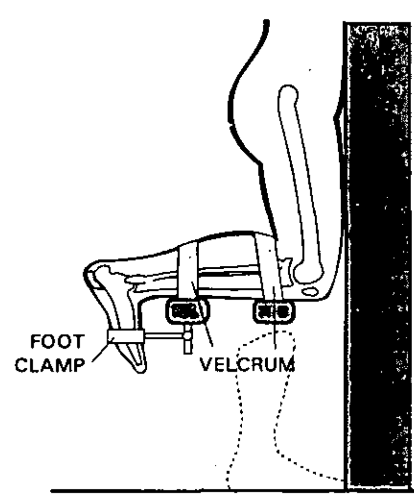


Figure 7.7. Details of leg and foot clamping

7.5.1.2. Impacting pendulum

The impacting pendulum consists of an **oscillating part** that is basically a cylindrical device of adjustable mass. This part was suspended from a structure mounted on the test rig called the **fixation head** that allowed adjustment of pendulum's position into the test rig.

The Oscillating part strikes the heel pad [Figure 7.8] and consists of a steel cylindrical rod on to which weights were mounted and secured by aluminium pieces screwed at each end.

The fixation head from a bar suspended the pendulum across the top of the test rig using four non-extensible nylon wires of equal length (147 cm). They were non-extensible to avoid possible errors in pendulum's movement.

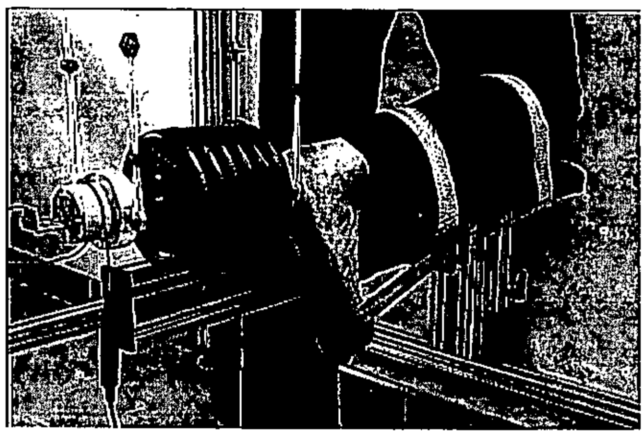


Figure 7.8. Oscillating mass

The wires were mounted in the pendulum by spherical joints arranged in the medial horizontal plane of the pendulum forming a rectangle allowing rotation and avoiding frictional losses. If the centre of masses of the pendulum is aligned with the oscillation axis then the pendulum movement will approach ideal pendulum movement. Wires also included positioning and security systems. The four-point suspension provided a parallelogram arrangement, which secured a circular translation of the impacting part, that is an horizontal and planar movement along the oscillation axis reducing vibration in pendulum after impact and ensuring an impact perpendicular to the sample being tested. The oscillating part consists of four elements (Figure 7.9).

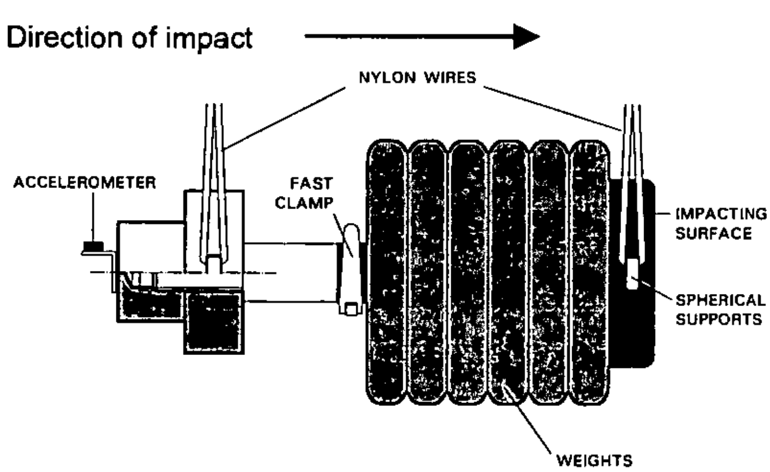


Figure 7.9. Oscillating part

The support rod is a steel rod into which weights were mounted to get the desired impacting mass. The rod and weights were commercially available fitness training apparatus. Two quick closing clamps were used to fix the weights on the central section of the rod to avoid capsizing the pendulum during oscillation. The total weight of the oscillating part without weights was 2.155 kg and the impacting mass could be adjusted from 2.155 kg up to 32.155 kg.

The impacting end is a cylindrical head of aluminium, 70 mm diameter, with rounded edges to avoid hurting the participant and screwed to one end of the support rod. The contact surface of area 38.48 cm^2 was polished for comfort and to ensure total contact with heel pad during the impact. This end incorporated two of the supporting spherical joints.

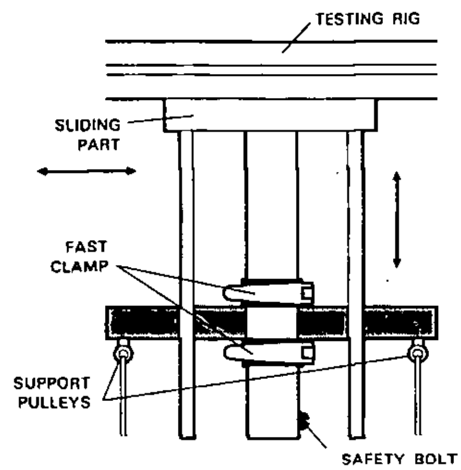


Figure 7.10. Fixation head

The rear end closing piece is a similar but smaller aluminium cylinder screwed at the rear end of the support rod. This piece had a double purpose: to hold in place all the elements of the oscillating mass and to serve as a support for the accelerometer.

The pendulum was suspended from **the fixation head** able to slide along the test rig. The head was designed to allow pendulum oscillation and adjustment in the test rig (Figure 7.10). The position of the pendulum can be adjusted both in the vertical and in the horizontal axis to ensure that the impacting mass would always be in a maximal horizontal velocity position at the moment of impact. The fixation head was mounted on the top bar of the test rig that can be moved sideways to fix the pendulum in the right position for impact. Part of the instrumentation for data triggering and drop height reading was mounted in this structure (Figure 7.11).

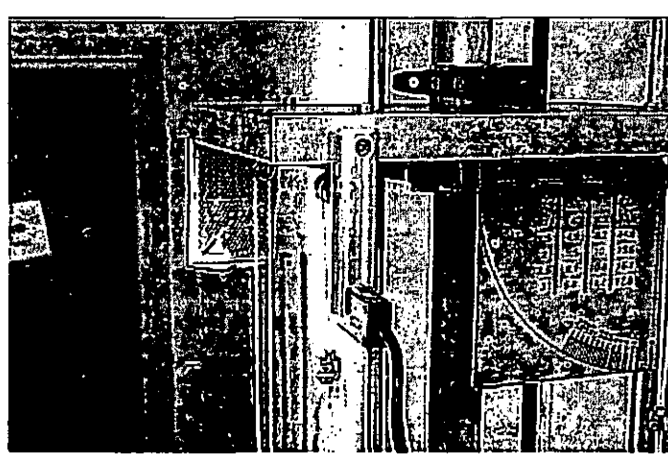


Figure 7.11. Instruments mounted on the fixation head

The fixation head consists of two parts and capable of adjustment horizontally by sliding-along the top bar and vertically (Figure 7.10). An aluminium cylindrical rod of 30 mm diameter and 300 mm long was fixed vertically in the centre of the sliding piece for vertical positioning of the oscillating mass. Two steel bars were fixed in this piece to avoid lateral movement and capsizing of the pendulum. At the lower end of the cylindrical rod a bolt was incorporated to secure the equipment and to avoid equipment damage in case of mechanical failure of any part (Figure 7.10).

The pendulum was suspended from an aluminium plate that can be moved up and down along the aluminium rod for vertical adjustment of the impacting mass. Four low friction pulleys were mounted in the lower side of this plate to support the impacting pendulum and arranged in a rectangle according to the position of the spherical joints in the oscillating part. Two quick clamps, below and above of this plate was used to fix this plate in position in the vertical rod.

7.5.2. Instrumentation

The testing procedure of the pendulum requires that the acceleration of the oscillating mass be registered during the impact to calculate the mechanical properties of the heel pad both directly and by double integration. Therefore, the instrumentation comprised all the measuring equipment for registering and storing the pendulum's acceleration (Figure 7.12). The instrumentation comprised an accelerometer, a Reflex photocell, acceleration amplifier, photocell electronics, a connecting box, a data acquisition card, one personal computer and a drop height measuring scale.

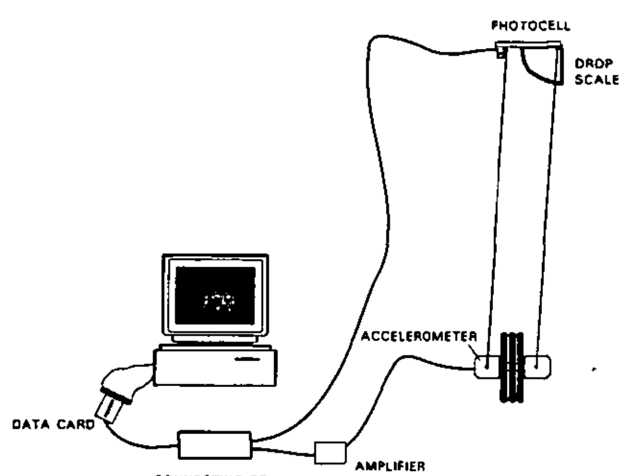


Figure 7.12. Instruments

The structure of the testing rig was suitable for fixing the electronic parts and the slots in the bars for the wiring.

7.5.2.1. Equipment for Registering Acceleration

The acceleration of the pendulum during impact was registered by means of a low mass uniaxial piezoresistive accelerometer from IC-Sensors [Model 3031]. This was mounted in the closing piece of the impacting mass by an aluminium frame built in two parts (Figure 7.13) to dequately align the accelerometer's axis with impact acceleration signal, i.e. the horizontal axis of the impacting mass.

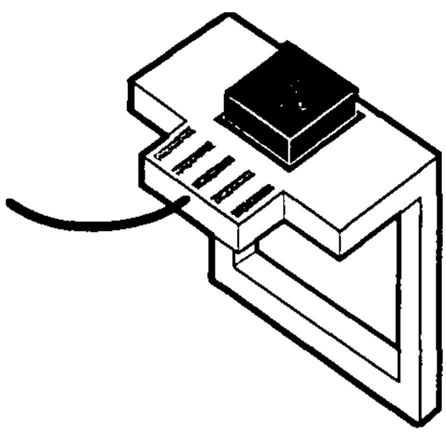


Figure 7.13. Accelerometer mounting

A piezoresistive accelerometer was preferred to a piezoelectric accelerometer because of its better response to low frequency signals, good linearity, low mass and reasonable price. The accelerometer used was uniaxial since the acceleration is always in the horizontal axis. The sensitive axis of the accelerometer was disposed parallel to the longitudinal axis of the pendulum in such a way that impact acceleration was positive. The accelerometer used for impact testing was mounted with the sensitive axis perpendicular to its surface with a total weight of 2.15 grams.

The signal from the accelerometer was transmitted to electronics developed at the IBV for amplification and for electrical conditioning. The accelerometer was connected to the amplifier by a wire in two stages via a connector that readily disconnected if violently pulled to prevent accelerometer damage.

The amplified signal was transmitted to a personal computer through a connecting box and a data acquisition card. The signal was registered in volts. The signal of the accelerometer had to be calibrated for conversion of volts into ms^{-2} (Appendix A2) for the calculations to be in the International System Units

7.5.2.2. Equipment for Data Triggering

The goal of this method is to study the impact, it is necessary to registering the acceleration only during the impact event which would also allow a higher sampling frequency with higher resolution. At the same time, this would avoid storing huge amounts of useless data thus reducing storing and testing time. Therefore, a method for acquiring data only during the impact was developed.

A reflex photocell was mounted for triggering data acquisition. This emits a light beam and receives the reflected beam. When the light barrier is interrupted an electric pulse is sent and data acquisition begins. The photocell was mounted in the lower plate of the fixation head with the reflector in front. A special device for mounting the photocell was designed. A small metal plate fixed to one of the nylon wires of the pendulum is used to disrupt the light beam to start data acquisition (Figure 7.11). After several trials its position was fixed to begin data acquisition a few milliseconds prior to impact.

The pulse from the photocell was sent to the data card in the computer to start data acquisition software by means of an electronic device that controlled the electric supply and differential voltage reference. Data acquisition end automatically after a time fixed by keyboard depending on the impact duration.

7.5.2.3. Equipment for connection with the computer

Accelerometer and photocell were plugged to a connecting box, which is an electronic device that sends the signals to the computer through a data card, which acts as interface with the computer. This device was designed to work with up to 8 analogue-digital output signals and up to 4 digital input signals. The output of the accelerometer is an analoge signal whilst photocell's output is digital (a pulse). A data acquisition Card PCMCIA from National Instruments was used for data transmission to the computer.

7.5.2.4. Drop height scale

An angular scale was mounted in the fixation head to control the height of drop of the pendulum (Figure 7.11). This scale consists of an adhesive graph fixed in a metallic piece screwed to the lower plate of the fixation head. The scale represents directly the drop height of the pendulum-rather than degrees of angular position. The position of the scale in the fixation head was chosen to ease the operator's task. The desired drop height was fixed by pulling the pendulum backwards until the corresponding mark of the scale was placed between the two branches of the corresponding nylon wire.

7.6. Software

Specific software was developed both for testing and for parameter calculation. The software developed can be divided into a program for **Data Acquisition** and for **Data Processing**. The former was intended for testing and obtaining data files which were stored for further processing by the latter which performed operations that process the parameters used for describing mechanical properties of the heel pad. Result were filed and stored for further statistical analysis.

7.6.1. Software for Data Acquisition

The software used for data acquisition was a version of a general data acquisition program developed at the IBV. This program was implemented as a stand-alone application in windows called PAVI.

This program was developed in LabWindows CVI for windows. It controls the data acquisition card and sampling parameters to register the acceleration signal during the impact. Tools for visually assessing the quality of the impact in the computer screen were included. Once the signal was acquired it was represented in volts along time in milliseconds.

PAVI included several facilities for data triggering, sampling set up, graphics and data saving. A description of general operation and features of the software is given in appendix A3, the specific application for impact testing is explained later.

7.6.2. Software for data processing

Software for data processing was developed in *Matlab* under windows. This software consists of a principal program named **Pendul1** and included general management and some specific signal operations and a series of functions developed for processing impact data. The software is listed and described in appendix A4.

A description of the software follows the sequence of data processing.

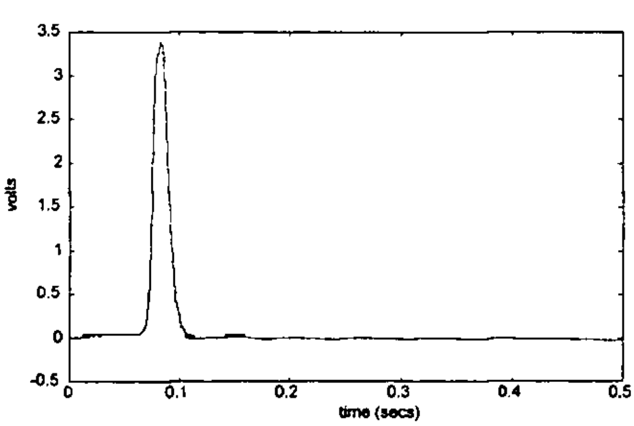


Figure 7.14. Amplified signal from accelerometer (volts).

The software is started by typing pendul1 in the Matlab command window which reads the impact signal (Figure 7.14) which is then reduced, filtered and converted to international units for further processing (Figure 7.15). A threshold value of 2.5% of peak was used to define start and end of impact for cutting. A cut-off frequency of 150 Hz was chosen considering that most of frequency content of heel strike during walking is below this value (Johnson, 1988; Smeathers, 1989).

Displacement at the moment of impact is set to zero since displacement to be studied is that of heel pad. Frequently, in the literature (Aerts et al., 1995; Valiant, 1984), impact velocity was calculated from potential energy change. In this work the velocity at impact was obtained for each test from velocity calibration curves obtained by integrating three acceleration signals from a free oscillation (no impact) of the pendulum from the testing drop height. Several functions were used to calculate the study parameters (Figure 7.16).

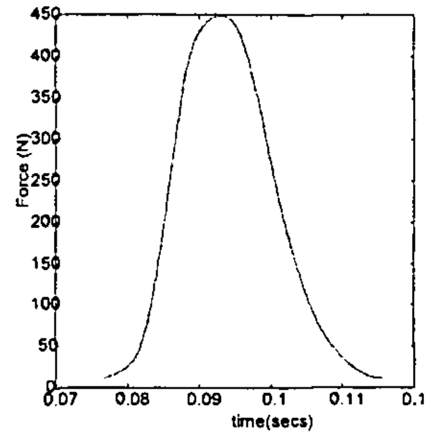


Figure 7.15. Force signal after preprocassing.

The energy absorbed is calculated as a percentage of input loading energy as follows:

Energy absorption = (Energy absorbed/Loading Energy) x 100

The energy is calculated as the area under the force-displacement curve and the software computes parameters from force-displacement curve

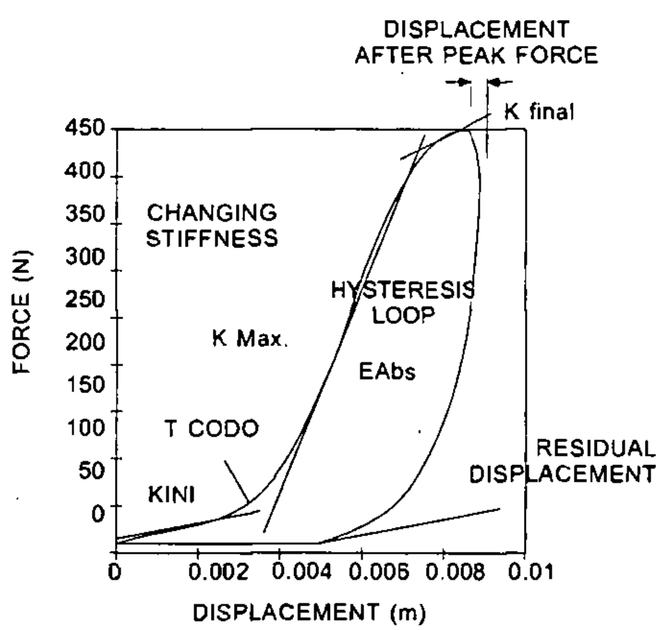


Figure 7.16. Force-displacement parameters

The impacting mass was a steel rod 80 cm long weighing 5 kg. The impacting area was circular of diameter of 40 mm (12.57 cm^2). An accelerometer and a displacement linear transducer (DLT) were mounted on the impacting mass.

Peak and time-to-peak are obtained from force and displacement. Stiffness corresponding at force peak (Kmean) is calculated as peak force divided by the corresponding displacement.

A software routine is called for parameters describing the load-displacement curve. As it can be observed in the load-displacement curve (Figure 7.16) - as in previous work - there is an initial low stiffness region followed by a high stiffness region and a final medium stiffness to peak force. In this sense, initial (Kini), maximal (Kmax) and final (Kfinal) stiffness for the loading phase as well as the time of shift from initial to maximal stiffness (Tcodo) are calculated as will be described later in this Chapter. Besides, heel pad displacement continues after maximum force has been reached and a residual displacement is observed at zero level force. Time delay between maximal force and maximal displacement is calculated (Figure 7.16).

7.7. Pendulum set-up

The pendulum set up procedure was as follows

1. Accelerometer calibration.
2. Assessment of the quality of pendulum movement and frictional losses.
3. Adjustment of pendulum mass and drop height to simulate walking heel strike.
4. Adjustment of photocell data triggering.
5. Calibration of the velocity curve.
6. Software checking.
7. Estimation of energy absorption due to the supporting wall.
8. Assessment of reliability and repeatability of the method.
9. Estimation of energy absorption due to movement of ankle and foot joints.

7.7.1. Accelerometer calibration

Accelerometer output is registered in volts, but for calculation purposes the acceleration must be converted to SI units (ms^{-2}). So the accelerometer was calibrated to obtain a relationship between the output in volts and the acceleration in ms^{-2}.

The procedure for accelerometer calibration was a general procedure used at the IBV. The calibration was done in the ± 1 g (gravity acceleration) range. The experimental set-up and results are included in appendix A2.

The calibration equation used for this accelerometer in ms^{-2} was:

$$Acel\ (g's) = 9.81 \times ((1.832 \times volts) - 0.0006)$$

7.7.2. Analysis of Pendulum movement

The pendulum must oscillate along a circular translation in a plane to ensure horizontal and central position with respect to the heel pad at impact, as well as to avoid energy losses due to vibrations, capsizing and others.

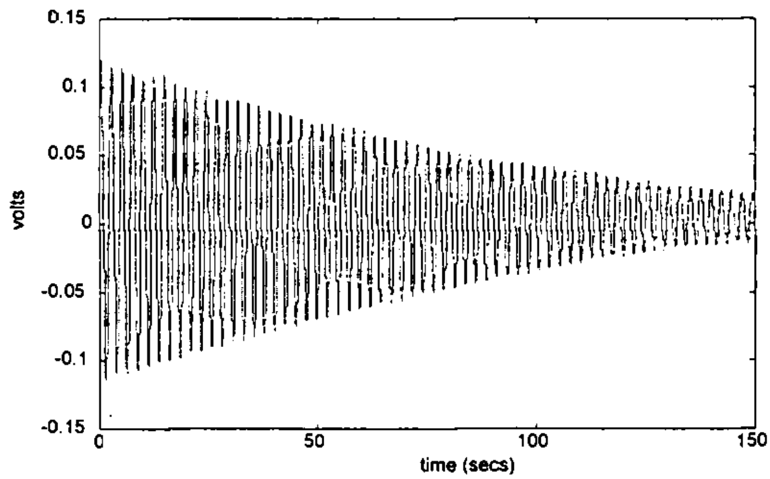


Figure 7.17. Signal from a free oscillation of the pendulum during 150 seconds.

After fixing the pendulum weight, the mass was regularly distributed into the oscillating mass and the length of the wires adjusted to ensure the horizontally of the impacting mass at rest position and checked using bubble levellers. The movement of pendulum was analysed during a free oscillation of 150 seconds, which is one thousand times longer than an impact (Figure 7.17).

Visual assessment revealed that movement occurred in a plane without capsizing. Acceleration during free oscillation showed an oscillating damped movement (Figure 7.17). The damping coefficient and natural frequency were calculated using a function (Sdof2.m) developed in Matlab for the analysis of three consecutive peaks to assess frictional losses (Appendix A4). The damping coefficient was found to be 0.29, which is rather low.

Comparison between the oscillation frequency of the ideal pendulum and the calculated frequency was used for assessing the quality of pendulum movement. The frequency of oscillation of an ideal pendulum is $fn = (1/2\pi) \times \sqrt{(g/l)}$, where l is the pendulum arm. In the device developed, the arm is 147 cm, so the ideal frequency would be 0.411 Hz. The frequency calculated was 0.42 Hz. The small differences were probably due to frictional losses and to the assumption made in the ideal pendulum, where all the mass is considered as located in a point at the end of the pendulum arm. These differences did not seem to have a relevant influence for impact studies.

7.7.3. Adjustment of pendulum mass and drop height

The pendulum was set up for a research project on shock absorption in adults during walking. A typical heel strike in adults barefoot walking gives a peak force around 0.8 of body weight in 15 ms, so impacting mass and drop height must be adjusted to deliver an impact energy of this order. On the other hand, according to the literature different combinations of mass and impact velocity (i.e. drop height) resulting in the same impact energy yield similar results. After several trials the pendulum mass was set to 8.155 kg and the drop height to 2 cm (theoretical impact velocity of 0.626 ms^{-1}), what resulted in an impact energy around 1.6 J and a loading history approximately similar to walking heel strike.

7.7.4. Adjustment of data triggering

The reflex photocell position and the location, shape and dimensions of the plate built for interrupting the light beam were chosen to adjust the start of data registering to just a few milliseconds prior to the impact. This adjustment was to increase sampling frequency (i.e. resolution) and to reduce, at the same time, the size of the data files to save computer memory and speed up data processing.

The minimum value for sampling frequency had to be of 7000 Hz per signal to avoid noise aliasing (greater than double of the maximum frequency considered as noise). In this application it was chosen to be minimum since this is enough for a good representation of the impact (it may be recalled that walking impacts have a frequency content lower than 150 Hz).

The number of data points to register has been selected according to the sampling frequency to get the whole peak. From the literature data an initial adjustment was for a peak duration around 50 ms (Folman et al., 1986; Jefferson et al., 1990; Jorgensen et al., 1989; Ligth et al., 1980; Shorten & Winslow, 1992) and then checked. Finally 500 miliseconds (tenfold of peak duration) of signal were collected. Data triggering was adjusted to be between 10 and 25 ms before impact for a drop height of 2 cm.

7.7.5. Calibration of initial velocity curve

In pendulum tests in the literature, unless there was direct measurement of either velocity or displacement, double integration of acceleration was usually used for determining the mechanical properties of the heel pad. In this sense, impact velocity and initial displacement were required as initial conditions for first and second integration respectively. Initial displacement was set to zero since it is the heel pad displacement that is of interest, but initial velocity needs more discussion.

In the literature this value is usually obtained from energy calculations as follows. The change in potential energy from drop position to the position at the moment of impact, PE = mgh which equals the kinematic energy at impact $Ec = 0.5mV_i^2$, then velocity at impact is $V_i = \sqrt{(2gh)}$. This value for a drop height of 2 cm would be 0.626 ms^{-1}.

To assess the relevance of a possible error in impact velocity estimation, impact signals for one participant were integrated using two different velocities as initial conditions, one the proposed test velocity and other 3.2% lower (0.62 vs. 0.6). With the higher initial velocity, energy absorption was underestimated by 6.3% and rigidity was overestimated by 4.7% (not significant) with respect to the lower. From these results the importance of a good estimation of initial velocity became clear

Other possible sources of error in this procedure were as follows:

- *Energy calculations* assume that pendulum is at the lowest point of its trajectory, so that horizontal acceleration is zero and horizontal velocity is maximum at the instant of impact. However, pendulum position at impact may be not always the same, since the position of the heel to test can change from participant to participant. Frictional losses should also be expected to reduce velocity.
- *Data collection* begins around 20 ms before impact that is before the impacting mass has reached the lowest point. Afterwards, for data processing the signal is restricted to only the impact part of the signal using a threshold value of 2.5% of peak. As a result, the initial moment of the acceleration signal used for integration will probably not correspond with maximum velocity instant. Until equipment for directly measuring either velocity of displacement is available, a procedure to calculate initial velocity for each testing session was used. The accelerometer signal during three free oscillations of the pendulum from the drop height was registered collecting pendulum's acceleration and a pulse from the photocell every time the pendulum crossed the light beam. Both signals were stored for further analysis (Figure 7.18).

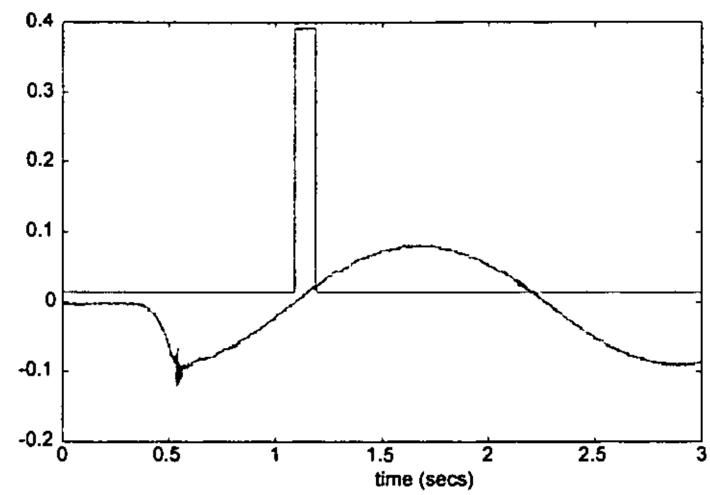


Figure 7.18. Acceleration and photocell signals (volts) for e drop height of 2 cm

The acceleration signal was processed using the same software and procedure developed for signal processing to obtain a velocity curve (Figure 7.19). The time lapse from the start of the oscillation to the rising flange of the photocell pulse was also stored as **calibration time** for further calculation.

To obtain the impact velocity to be applied in data processing, the calibration time has to be added to the time elapsed in the test from data triggering until force signal reaches the fixed start threshold (2.5 % in this case). Then the velocity in the calibration curve corresponding to the resulting time will be the velocity to be used for integration.

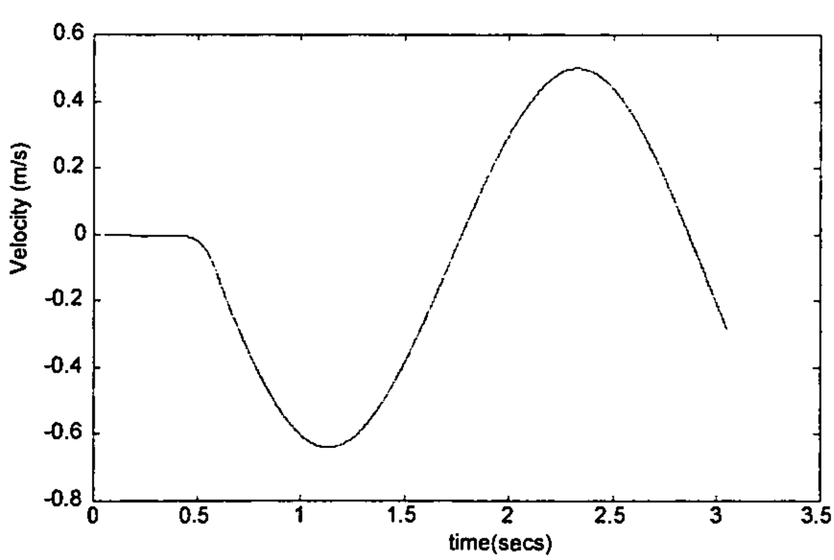


Figure 7.19. Curve velocity for e drop height of 2 cm.

Nevertheless, this procedure does not avoid errors due to inaccurate positioning of the pendulum at the corresponding drop height. In a general sense, measuring displacement would be preferred rather than using double integration and this was given further consideration.

7.7.6. Software checking

For checking the integration function, a sinus signal was integrated and the result compared with a minus cosine signal ($\int \sin x dx = -\cos x$). No errors were detected and both theoretical and calculated signals were very similar.
The whole program was also checked in different trials for assessing reliability and repeatability of the method. No errors were observed. The method will be explained later.

7.7.7. Estimation of contribution of the supporting wall

The participant is instructed to firmly press the knee against the supporting wall. As reported in the literature (Aerts & De Clercq, 1993c) the supporting wall must have a high rigidity to avoid small vibrations which would absorb energy producing an overestimation of energy absorption.
In the procedure developed in this work, a thick and heavy leaded wall (formally used for X-ray screening) was used to minimise the aforementioned effect. Nevertheless, to assess wall's contribution an accelerometer was mounted on the wall to registering its acceleration during human testing. Comparison of both signals was used to assess if any relevant vibration occurred. A signal from the accelerometer mounted on the wall recorded during a human testing was analysed and no resonance vibration was observed, thus wall's contribution was assumed to be negligible.
Testing of a steel spring of known stiffness constant was used for assessing the reliability and repeatability of the method, which also allowed analysis of wall's contribution. Any energy loss or differences in the spring stiffness obtained should be due to wall vibrations.

7.7.8. Analysis of filtering frequency

Spectral analysis of pendulum acceleration using Fast Fourier Transform was done to establish the frequency needed for filtering the signal to avoid noise, wall vibrations and others effects, controlling - at the same time - possible loss of information due to filtering (Appendix A5). The analysis of the spectra power revealed that more than 90% of information was contained in the part of the signal below 150 Hz.

7.7.9. Assessment of reliability and repeatability of the method

Pendulum testing has been shown to be reliable and repeatable by different authors (Aerts et al., 1995; Cavanagh et al., 1984; Valiant, 1984). Repeatability has been assessed as 1 - Coefficient of variation (where COV = standard deviation -mean). Values found in the literature are around 0.96 for heel pad deformation, 0.95 for compression energy and 0.85 for decompression energy (Jorgensen et al., 1989a).

To assess reliability and repeatability of the method several tests were carried out. Initially, three valid trials were performed on ten participants (5 male and 5 female) and the average repeatability (1- COV) was calculated for energy absorption. A high repeatability of the method was found (0.927). At the same time, energy absorption ranged from 65% to 85%, mean value of 73%, which agrees with results found in the literature.

Spring test

A commercial spring made from chromium-vanadium steel (DIN 17225) was mounted on the wall to assess the reliability of the procedure. Mounting was problematical so a double-sided adhesive tape was used. As a consequence, the unloading phase was not very reliable, so energy absorption results were not considered. Stiffness was then used for validation. Stiffness as given by manufacturer was 109 ± 10% kN/m, value calculated from pendulum was 98.8 kN/m what represent an error of (109 – 98.8)/109 = 9.3%, lower than manufacturer values. The repeatability was very high (1-COV = 0.91) for stiffness (Table 7.1).

Table 7.1. Spring test results

	Stiffness (kN/m)	Energy absorption (%)
Mean	98.8	36.3
Standard error	2.7	2.36
Standard deviation	8.5	7.47
Confidence level (95.000%)	5.3	4.6
COV (sd/mean)	0.086	0.21

Energy absorption was not considered due to methodological problems. Note that repeatability (1 – 0.21) was as low as 0.79 so this test should be repeated for energy absorption with a better method for fixing the spring to the wall.

7.7.10. Considerations on the energy absorption due to the movement of the ankle and foot joints

The pendulum test is aimed at studying passive shock absorption of the lower leg, so any contribution from active mechanisms should be avoided. Given the participant's position for testing, active shock absorption could occur from muscle tension and joint movement.

Participants were instructed to relax their leg muscles during the test but the forced position of the knee against the wall and positioning the foot in dorsal flexion will tense the thigh and calf muscles and probably prevent most of muscle absorption. Impact was performed without warning to the participant to avoid non voluntary muscle reaction.

Joint movement must, however, be prevented by mechanical restriction. The knee joint was fixed in flexion by participant's position so its contribution can be neglected. Ankle and subtalar joint of the foot were supposed to help shock absorption during walking. At heel strike during walking, the foot is slightly dorsiflexed and supinated, following sudden pronation and plantar flexion are regarded as active shock absorbing mechanisms. This angular position of the foot has been used by some workers for drop and pendulum heel pad testing. However, a great inter-individual variability in foot joints' range of mobility and flexibility is observed thus a fixed ankle position would represent different relative joint clamping which could leave some active absorption for some participants. So for each participant maximum angle dorsiflexion by locking both ankle and subtalar joints instead of fixed angle dorsiflexion was preferred as this eliminated joint shock absorption.

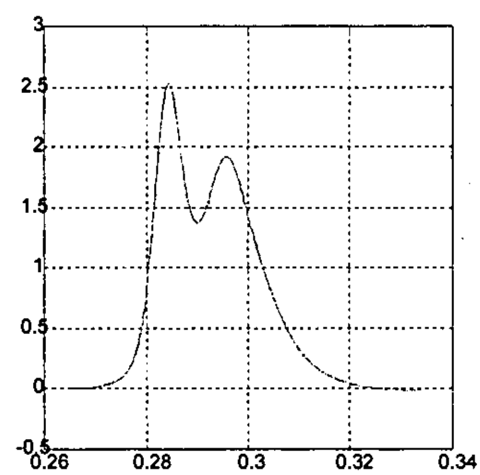


Figure 7.20. Double peak impact signal due to ankle mobility

To assess this effect, a participant with the ankle slightly dorsiflexed - as described in the literature - was tested. Results showed a double peak acceleration signal (Figure 7.20) as reported by Valiant (1984) who largely disregarded these signals. Results from this signal showed higher shock absorption (greater than 95%), higher maximal and residual displacements (probably due to joint displacement) and lower peak force (Figure 7.21) than the same participant tested at maximal ankle dorsiflexion. Force-displacement curves showed an unusual pattern (Figure 7.21). In this sense, double peak signals were considered to indicate bad joint clamping and excluded of the analysis.

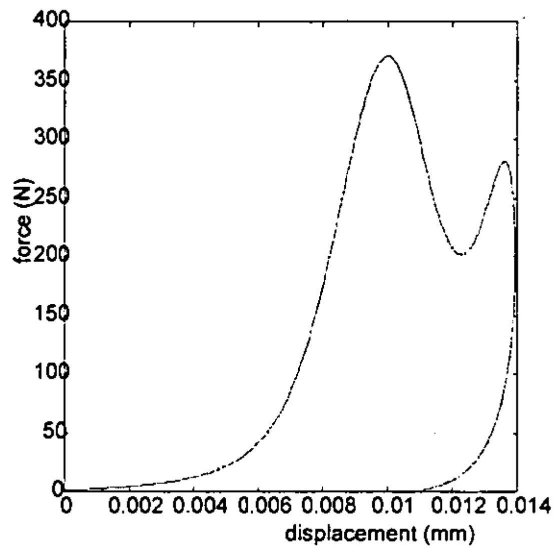


Figure 7.21. Force-displacement showing enkle ebsorption

7.8. Testing Procedure

Figure 7.22. Testing set up

The testing procedure was designed to reduce as much as possible the time spent by the participant in the test rig. It consisted of:

i. Connection and pendulum set up.
ii. Participant characterisation.
iii. Adjustment of test rig.
iv. Testing

All the equipment must be correctly connected at least ten minutes allowed for warming up. It is very important to switch on the computer before inserting the data card to avoid damage from any voltage peak. All equipment is then switched on and the Datacard inserted in the PC socket with the computer on. Windows is then started and then the acquisition program PAVI (for configuration and software controls see earlier this chapter and appendices) followed by selection of the required configuration including sampling frequency, number of channels and time of acquisition. Select data triggering mode to keyboard.

7.9. Methodology for the analysis of heel pad mechanics and of the influence of individual's characteristics

A total of 54 healthy volunteers participated in this experiment. Eighteen active elderly (age range 51-81) and thirty-six healthy young adults (age range 18-30) were selected and divided in four analysis groups: 18 male adults, 18 female adults, 9 active elderly male and 9 active elderly female. Weight, height, age and gender data were collected from each participant and stored for further analysis. Physical and medical records were also collected. The characteristics of each group are shown in Table 7.2.

Table 7.2. Participants' characteristics.

			ADULTS		
	Mean	Standard Deviation	Median	Maximum	Minimum
FEMALE					
BMI	21.85	1.96	21.37	27.59	19.05
AGE	22.67	3.55	22.00	28.00	18.00
HEIGHT (cm)	159.44	4.31	160.00	168.00	151.00
WEIGHT (kg)	55.56	5.32	55.00	68.00	48.00
MALE					
BMI	24.08	2.83	23.54	31.44	19.69
AGE	23.50	2.50	23.50	29.00	19.00
HEIGHT (cm)	173.67	5.67	172.50	187.00	164.00
WEIGHT (kg)	72.61	9.00	70.75	93.00	60.00
			ELDERLY		
FEMALE					
BMI	29.06	4.73	27.11	36.12	23.50
AGE	57.22	6.80	56.00	73.00	51.00
HEIGHT (cm)	151.00	4.30	152.00	158.00	145.00
WEIGHT (kg)	66.23	10.85	64.40	81.70	55.00
MALE					
BMI	27.86	4.17	28.05	35.11	23.05
AGE	69.33	7.42	70.00	81.00	59.00
HEIGHT (cm)	168.22	7.50	169.00	185.00	161.00
WEIGHT (kg)	78.89	12.85	81.00	96.00	61.00

All participants were informed about the study and agreed in writing to participate. The Body Mass Index (BMI) was used for estimating obesity. The BMI is calculated as:

$$\text{BMI} = \text{Weight in kg} / (\text{height in metres})^2$$

A participant with a BMI greater than 27 was considered to be overweight. Thirteen participants presented overweight (BMI > 27), ten elderly (five male and five female) and three young (one female and two male). BMI has been chosen as a simple method of estimating fat body content so to study its relationship with heel pad properties.

The ballistic pendulum developed for *in-vitro* testing of the heel pad was used. The test rig was adjusted in such a way that the participant stood comfortably on one leg, shod. The test leg was fixed to the wooden leg supports one of which was placed one under the knee, but touching the wall. The other support was placed under the tibia sufficiently far from the ankle to avoid contact with the pretibial muscles because this could give rise to vibrations in the signal that participants might find painful.

The leg was fixed to the supports by strips of Velcro® tightly bound but without pain, and the foot was clamped in dorsiflexion before adjusting pendulum. The ankle dorsiflexed, which was fixed at maximal dorsiflexion to avoid shock absorption due to joint mobility. The participant was placed in the test rig and impacting pendulum's position at rest was fixed to touch the heel pad. The participant was instructed to relax in order to avoid muscle absorption and involuntary muscle reaction and the impactor released without warning. For testing, the pendulum was pulled backwards to the desired drop height. The *Adquirir* routine was started and the pendulum released.

The validity of every trial was first visually assessed at impact and then analysed graphically in the computer screen. Only heel centred, mass horizontal and single peaked impacts were accepted. Three valid impacts were collected for each participant and stored for further analysis.

Data files were stored during a testing session. On completion of testing and data processing, a file with results for each participant was obtained. Then, all the result files of all participants were combined using Excel (Microsoft) and individual's characteristics were added in the same file. Some parameters such as BMI, delay times, etc. were calculated and included in the file. The final file was exported to SPSS 7.5.2.s and Statgraphics Plus 2.1 for windows for further statistical analysis.

Twelve parameters were calculated for describing heel pad behaviour under impact loading. Most of the parameters for the study were obtained from the data processing software; others were calculated afterwards. They can be divided into impact parameters and force-displacement parameters.

Impact parameters (Figure 7.23)

- Peak force (Fimpact) which was calculated as acceleration times impacting mass, (F=ma).
- Time to peak force (TFimpact).
- Time of impact duration (Tdura).
- Peak displacement (Dmax).
- Time to peak displacement (Tdmax).
- Time delay between peak force and peak displacement (Tvisco = TFimpact - Tdmax).

Force-Displacement parameters (Figure 7.24)

- Energy absorption (Eabs). Calculated as the loop area by integrating force-displacement curve as Eabs = (Loading Energy-Unloading energy) x100/ loading energy.
- Stiffness for peak force (Kmean). Obtained as Fimpact/Displacement at Fimpact.
- Maximum Stiffness (Kmax). Calculated as the maximum of the curve of stiffness (differentiation of Force with respect to displacement).
- Initial Stiffness (Kini). Obtained by linear regression in the region of the stiffness curve before the maximum.
- Final stiffness (Kfinal). Obtained by linear regression in the region of the stiffness curve after the maximum stiffness till the peak force.
- Time of shift from initial to maximal stiffness region (Tcodo)

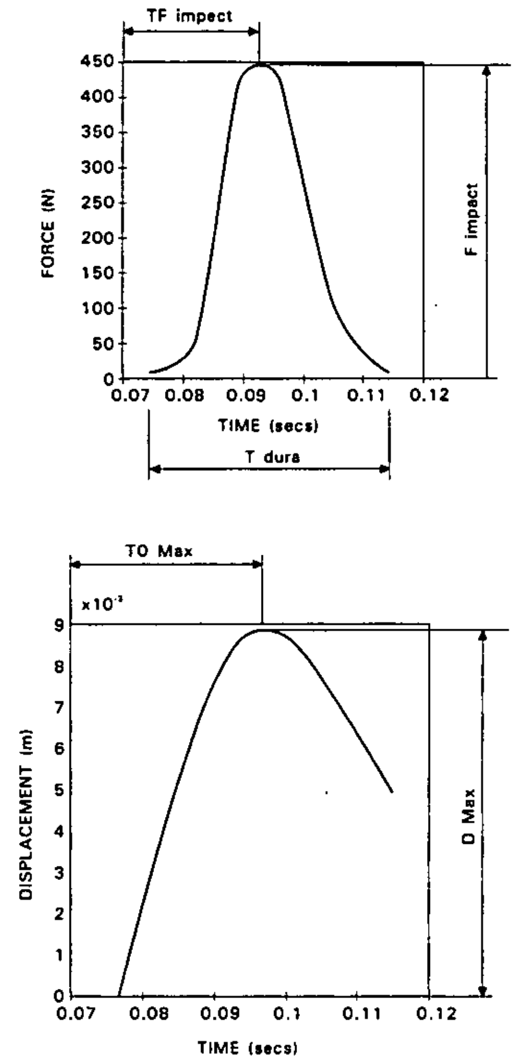


Figure 7.23. Impact paremeters

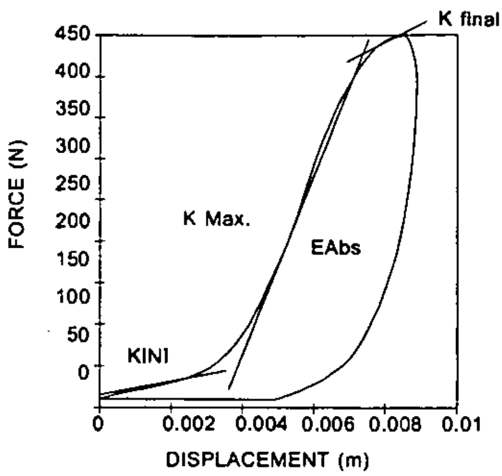


Figure 7.24. Force-displacement parameters

International units were used. Force was registered in Newtons, displacement in meters, stiffness in Newtons/meter and time in seconds. Descriptive statistics for all the parameters was done for each group using SPSS 7.5.2.s for Windows.

Peak force (FImpact), time-to-peak force (TFimpact), peak displacement (Dmax), time-to-peak displacement, (Tdmax), energy absorption (Eabs), maximal stiffness (Kmax) and initial stiffness (Kini) were studied as considered basic properties and for comparison with the literature results. Time delay between peak force and peak displacement (Tvisco) was studied with respect to the relationship between force and displacement after peak force which has been used in the frequency domain to measure energy absorption of viscoelastic materials (García et al., 1994). Time of shift from initial stiffness region to maximal stiffness part (TCodo) was calculated to analyse duration of low stiffness displacement. Final stiffness (Kfinal) was computed to study bottoming out. Stiffness at peak force (Kmean) was used to describe relationship between force and displacement at peak force, whilst time of duration (Tdura) was included to study the changes in the time taken to transfer the momentum during the impact, which is a mechanism of impact force reduction.

Factor analysis of Principal components was done for all the heel pad properties to better understand the complex relationship within variables eliminating false interpretations due to random and internal correlation between variables. Principal components method with Varimax rotation was used for extracting factors with the criteria of explaining a global variance greater than 90%. Factors with an eigenvalue greater than 0.6 were extracted which was considered valid in case that eigenvalue after rotation was not lower than 1 (variance explained by a factor is not lower than that explained by a single variable). Varimax rotation was chosen because it makes easier to explain the meaning of each component since the contribution of variables to components is maximised. Variable's contribution to factors lower than 0.5 were not considered. SPSS 7.5.2.s for windows was used.

The influence of gender, age and overweight in heel pad components was analysed by Multifactor Analysis of Variance (ANOVA) considering gender, age group and obesity as factors. Statgraphics Plus 2.1 for windows was used. Differences were considered statistically significant for $p < 0.05$. LSD post hoc multiple range test was used to establish homogenous groups. A test power study was done for all the variables investigated, to evaluate the probability of finding statistically significant differences of a given size. Normality and homogeneity of variance were checked by Leben and K-s tests respectively. The variables are not independent but as the interest of the study was in measured and meaningful rather than in combined parameters Manova was not used.

Pearson's correlation analysis of components with impact forces and time-to-peak was finally done for each group, as well as for young and elderly separately to study age and gender influence in heel pad mechanics. Overweight influence was also studied by comparing correlation analysis for overweight elderly and for non-overweight elderly.

A test power study was done for all the correlation analysis performed to calculate the smallest statistically significant correlation likely to be found at 80%-90% power in each analysis. This analysis issued a critical value to evaluate the relevance of the differences in correlation found for each group. A difference was considered relevant when a significant correlation found in a given analysis was greater than the critical value for another group were no significant correlation was found.

7.10. Results

Table 7.3 shows descriptive statistics of the different parameters for young and elderly, both male and female.

Results for young showed that impact forces ranged from 426.16 N to 682.7 N, rising in a time from 10 ms to 17 ms, and lasting for an average of 40 ms. Average energy absorption was around 80% (from 75% to 91%). Mean peak displacement was around 7 mm (from 5.5 mm to 10.4 mm). Maximal stiffness ranged from 79.61 kNm^{-1} to 129.20 kNm^{-1} and initial stiffness ranged from 13.41 kNm^{-1} to 68.03 kNm^{-1}.

Results for elderly showed impact forces ranging from 388.1 N to 714.28 N, rising in a time from 10.26 ms to 19.62 ms, and lasting for an average of 42 ms. Average energy absorption was around 80% (from 75.6% to 93%). Mean peak displacement was 8.5 mm (from 5.6 mm to 11.2 mm). Maximal stiffness ranged from 59.53 kNm^{-1} to 180.77 kNm^{-1} and initial stiffness ranged from 12.21 kNm^{-1} to 63.68 kNm^{-1}. Although the results were quite similar to those for young, a greater variation was observed, mainly due to elderly male participants. Force-time and Force-displacement curves showed common patterns for all participants from all groups.

Table 7.3. Results for younger and elderly, both male and female.

	YOUNGER				ELDERLY			
	Mean	Standard Deviation	95% CI for the mean lower........upper		Mean	Standard Deviation	95% CI for the mean lower	upper
FEMALE								
Dmax (m)	.0074	.0012	.0068	.0080	.0086	.0015	0.0077	0.0095
Eabs (%)	82.19	5.41	79.58	84.8	82.56	4.95	78.87	86.25
Fimpact (N)	557.75	70.30	5263.29	592.22	532.40	85.08	483.65	581.14
Kfinal (kNm^{-1})	89.85	27.90	76.16	103.55	77.44	23.51	58.07	96.18
Kini (kNm^{-1})	38.00	13.60	31.52	44.45	27.73	12.14	18.56	36.90
Kmax (kNm^{-1})	119.54	25.41	107.53	131.55	105.56	31.38	88.58	122.55
Kmean (kNm^{-1})	82.96	23.10	72.62	93.30	68.06	22.61	53.43	82.68
Tcodo (Sec.)	.0028	.0013	.0019	.0036	.0040	.0015	.0029	.0052
Tdmax (sec)	.0165	.0024	.0152	.0177	.0187	.0031	.0169	.0204
Tdura (sec)	.0405	.0039	.0384	.0427	.0421	.0051	.0390	.0451
Tfimpact (sec)	.0133	.0022	.0122	.0144	.0154	.0024	.0138	.0169
TVisco (sec)	.0032	.0008	.0026	.0038	.0033	.0010	.0025	.0041
MALE								
Dmax (m)	.0069	.0012	.0063	.0095	.0085	.0017	0.0076	0.0094
Eabs (%)	84.64	6.19	82.03	87.25	83.84	4.68	80.15	87.54
Fimpact (N)	535.21	53.77	500.74	569.76	539.14	96.13	490.40	587.89
Kfinal (kNm^{-1})	85.00	30.11	71.30	98.70	80.01	33.14	60.64	99.38
Kini (kNm^{-1})	45.00	13.89	38.51	51.48	30.61	14.91	21.43	39.78
Kmax (kNm^{-1})	129.20	15.56	117.19	141.21	119.03	33.93	102.05	136.02
Kmean (kNm^{-1})	87.34	158.91	77.00	97.68	72.14	28.29	57.52	86.76
Tcodo (sec)	.0028	.0021	.0020	.0036	.0046	.0017	.0035	.0058
Tdamx (sec)	.0161	.0020	.0148	.0173	.0185	.0035	.0157	.0202
Tdura (sec)	.04031	.00336	.03818	.04244	.04367	.00663	.04066	.04669
Tfimpact (sec)	.0117	.0022	.0106	.0129	.0146	.0029	.0130	.0162
Tvisco (sec)	.0043	.0016	.0038	.0049	.0038	.0014	.0030	.0047

7.10.1. Principal Component Analysis

Three components accounting for 93.08% of total variance were extracted from Factor Analysis of Principal components (Table 7.4). The lower communality (variance of a variable explained by components extracted) observed was for final stiffness (Kfinal, 0.87) which was high. Components of each factor after Varimax Rotation are shown in Table 7.4. Rotated factors were extracted for analysis.

Table 7.4. Principal componantanalysis

FACTORS	Factor 1	Factor 2	Factor 3
VARIABLE			
Dmax	-0.531	0.832	
Eabs			0.958
Kfinal	0.855		
Kini	0.520	-0.800	
Kmax	0.897		
Kmean	0.734	-0.657	
Tcodo		0.952	
Tdmax	-0.632	0.740	
Tvisco			0.918
Tdura	-0.649	0.646	
Eiganvalue Before Rotation	6.783	1.878	0.647
% Explained Veriance	67.83	18.78	6.47
% Cumulative Variance	67.83	86.61	93.08
Eigenvalue After rotation	3.826	3.541	1.941
% Explained Variance	38.26	35.41	19.41
% Cumulative Variance	38.26	73.67	93.08

Components 1 and 2 accounted for more variance than component 3 as they had greater eigenvalues (3.83 and 3.54 respectively) and both together accounted for 73.67% of variance.

Component 1 was mainly and positively defined by all stiffness parameters, specially by maximal (0.90), final (0.86) and mean (0.73) stiffness, also lower positive contribution of initial stiffness (Kini, 0.52) was observed and low negative contribution of peak displacement (-0.53), duration time (-0.65) and time-to-peak displacement (-0.63) were also observed. This component increases as stiffness parameter increase and displacement and time parameter decrease, which could be related to elasticity mechanism for reducing impact forces due to the elastic nature of the heel pad. So it was called **elastic component**.

Component 2 result was an interesting feature of the analysis. It was mainly defined by a positive contribution of shift time (Tcodo, 0.95) and peak displacement (0.83) and lower contribution of time-to-peak displacement (0.74) and duration time (Tdura, 0.65), negative contribution of initial (-0.80) and mean stiffness (-0.66) was also observed. This component appeared to be related to elastic properties at initial loading and duration of this phase, and to the peak displacement describing (a great) deformation at low stiffness, it was called **initial deformation component**.

Component 3 (eigenvalue of 1.94) accounted only for 19.44% of total variance. It was positively described by energy absorption and delay time between peak force and peak displacement (Tvisco), parameters related to viscoelastic properties of heel pad, so it was called **viscoelastic component**.

Thus, according to the theory of viscoelastic materials, the heel pad behaviour under impact could be described by the combination of an initial deformation component, an elastic and a viscoelastic component. Most of the variance (73.67%) of the heel pad properties was accounted for by elastic and initial deformation components. Thus the main differences in participants' heel pad properties are probably due to differences in structural factors affecting elastic systems.

Some confusion can appear when analysing the results with the nomenclature of the elastic component. It was chosen in relation to mechanical nature in contrast to viscoelasticity. In any case, it has to be born in mind that greater values of the elastic component correspond to a stiffer heel pad, which would help in understanding the results.

Impact mechanics of heel pad

Force and displacement curves obtained in this study are consistent with those showed by different authors (Cavanagh et al., 1984; Jorgensen & Bojsen-Moeller, 1989a; 1990). The analysis of the force-displacement curve showed non linear and viscoelastic features of heel pad under impact loading as in the literature (1984Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1993a, b; Valiant,). Displacement continued after peak force had been reached. In this sense, a positive time delay between peak displacement and peak force (Tvisco) was observed (3.62 ms ± 1.6 ms), which has been related to the viscoelastic nature of the heel pad (Valiant, 1984). After unloading, a residual deformation was observed at zero force level (Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1993; Valiant, 1984).

The elastic systems of the heel pad played a major role in impact mechanics. Pearson's Correlation analysis for all the participants showed high positive correlation of impact forces with the elastic component (greater elastic component means increased stiffness) (0.847) and a low negative correlation with the viscoelastic component (-0.420) (Table 7.5). Initial deformation component showed no statistically significant correlation with Impact forces. However, it also showed a high positive (0.781) correlation with time-to-peak force: greater times for lower initial stiffness and greater peak displacement and shift time, while elastic component showed a lower negative correlation with it (-0.529) (Table 7.5). Elastic systems acting at initial loading

(Initial deformation component) would mainly govern the time-to-peak force whilst the other elastic systems (Elastic component) would act on the peak force.

Table 7.5. Pearson's Correlation between elastic and viscoelastic components and peak force and time-to-peak (P = 0.05).

	Fimpact	TFimpact
Elastic component	0.847	-0.529
Viscoelastic component	-0.420	n.s. < 0.385 *
Initial deformation component	n.s. < 0.385 *	0.781

* indicates that a non significant (n.s.) correlation has been found and that in any case, according to the power of test of the experiment, the value shown would be the smallest significant correlation likely to be found.

7.10.2. Influence of physical characteristics in heel pad properties

7.10.2.1. Influence of age in heel pad properties

The elderly presented a longer time (0.0154 ± 0.0026 s) to force peak than the young (0.0125 ± 0,0024 s) (Figure 7.25). Peak displacement was greater in elderly (8.8 ± 1.6 mm) than in young (7.2 ± 1.3 mm) (Figure 7.26) and time-to-peak displacement (Figure 7.27) was also longer in the elderly (0.0186 ± 0.003 s) than in the young (0.0163 ± 0.002 s).

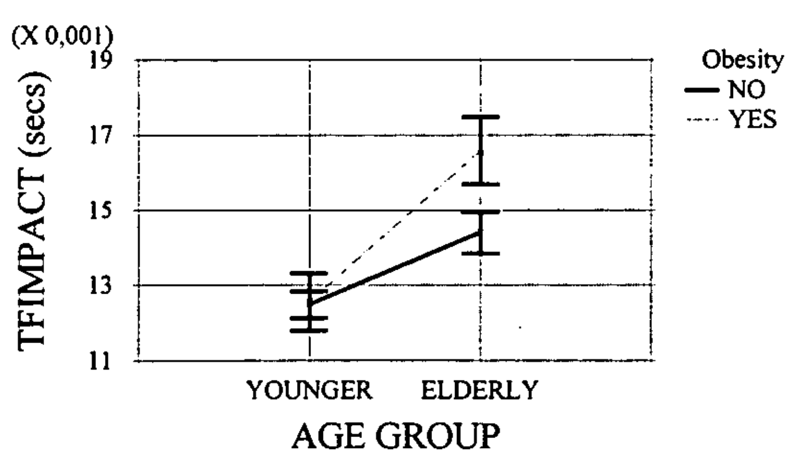


Figure 7.25. Interaction overweight-age group for time-to-peak force.

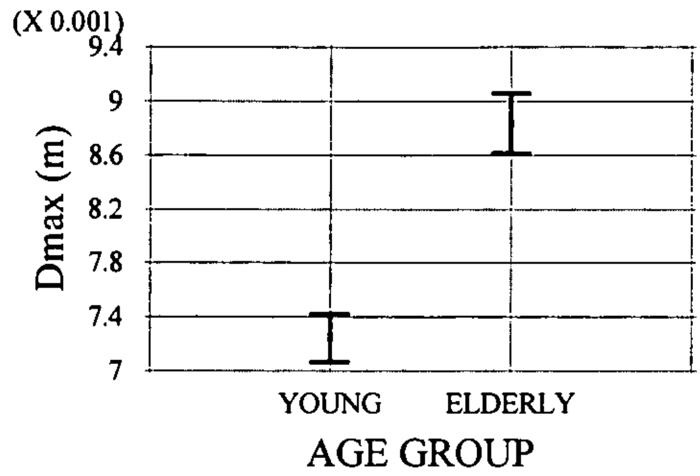


Figure 7.26. Maximal displacement for the younger end the elderly

Initial stiffness (Figure 7.28) and final stiffness (Figure 7.29) were greater for the young (respectively 41.4 ± 15.1 kNm^{-1} and 88.9 ± 30.4 kNm^{-1}) than for the elderly (29.2 ± 13.4 kNm^{-1} and 78.7 ± 28.7 kNm^{-1}). The time of shift from initial to maximal stiffness (Tcodo) was lower in the young (0.0028 ± 0.002 s) than in the elderly (0.0043 ± 0.0016 s) (Figure 7.30). No statistically significant differences were found for energy absorption.

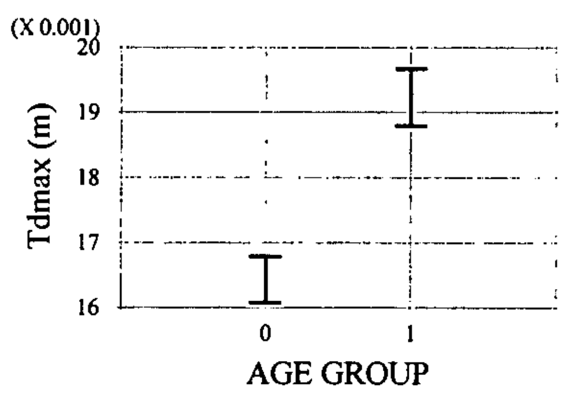


Figure 7.27. Time-to-peek displacement for the younger (0) and the elderly (1)

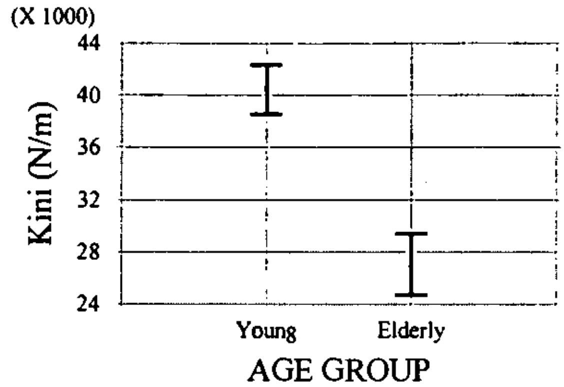


Figure 7.28. Initial stiffness for the younger end the elderly

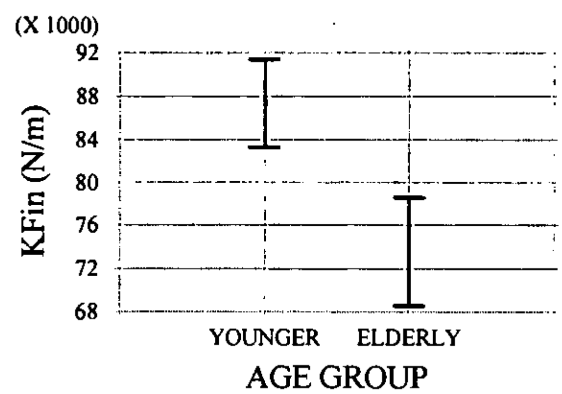


Figure 7.29. Final stiffness for the younger end the elderly

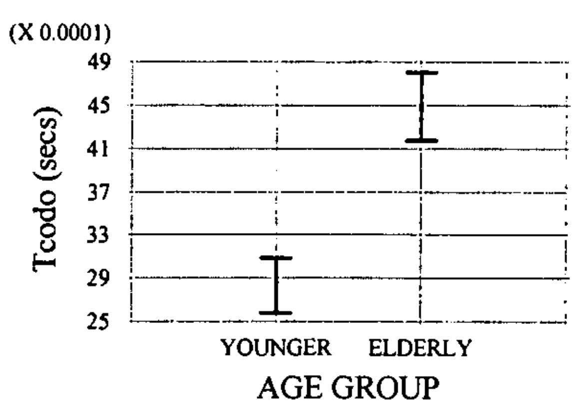


Figure 7.30. Shift time for the younger end the elderly

7.10.2.2. Influence of obesity in heel pad properties

Significant interaction was observed between age group and the overweight for impact force, time-to-peak force, time-to-peak displacement, energy absorption and maximal stiffness. Nevertheless, few overweight young (n= 3) were found in the participants (n= 36), since obesity is more frequent in the elderly. As a result, in this study, overweight was only analysed in the elderly.

Time to force peak was longer for the overweight elderly (0.016 ± 0.0025 s) than for both the non-overweight (0.0147 ± 0.0026 s) and the young (Figure 7.25). Overweight elderly showed impact forces (508.9 ± 86.1 N) statistically significantly lower than the young (548.49 ± 67.81 N) and the non-overweight elderly (569.02 ± 88.18 N). (Figure 7.31).

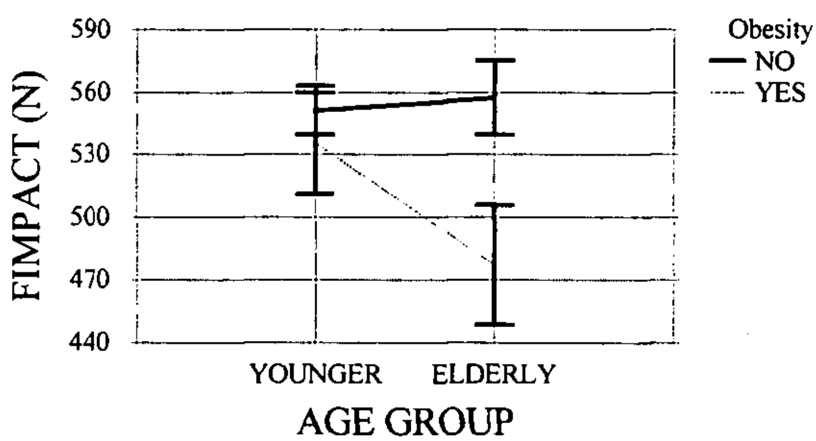


Figure 7.31. Interaction overweight-age group for impact forces

Peak displacement was greater in the overweight elderly (9.1 ± 1.5 mm) than in the non-overweight elderly (8.2 ± 1.4 mm) and young (Figure 7.32). Time-to-peak displacement was longer for overweight elderly (0.0196 ± 0.003 s) than for non-overweight (0.0182 ± 0.002 s) (Figure 7.33). Overweight elderly presented greater energy absorption (83.62 ± 5.52%) than non-overweight (81.00 ± 5.52%), but with no differences with younger.

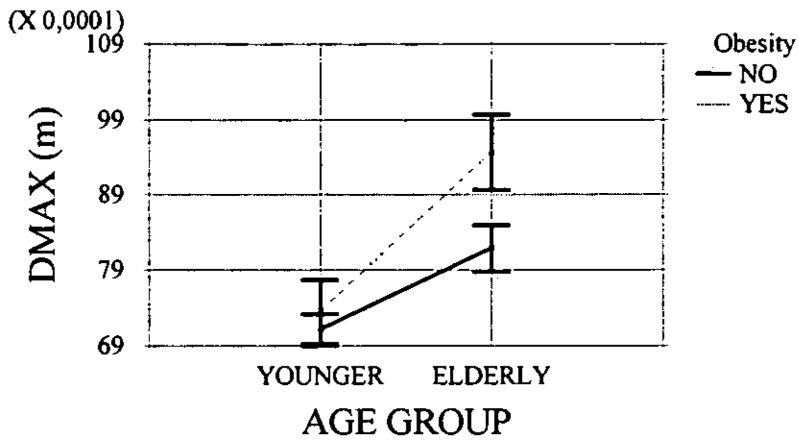


Figure 7.32. Interaction overweight-ege group for peak displacement

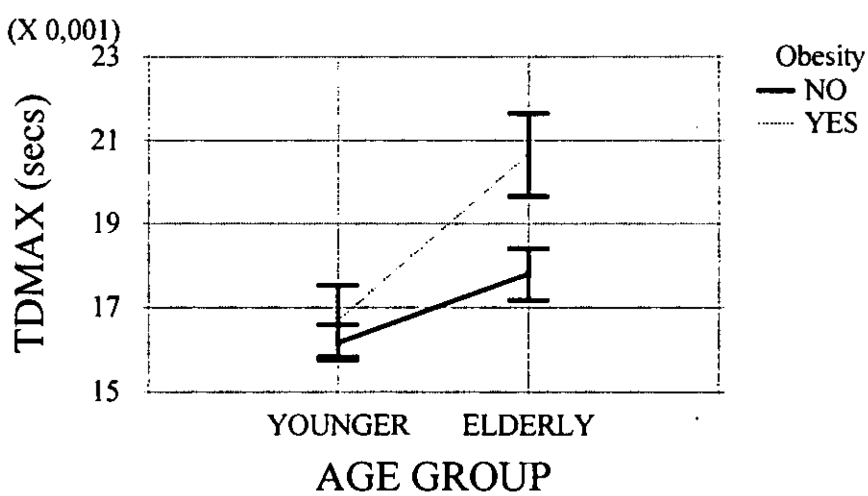


Figure 7.33. Interaction overweight-ege group for time-to-peak displecement

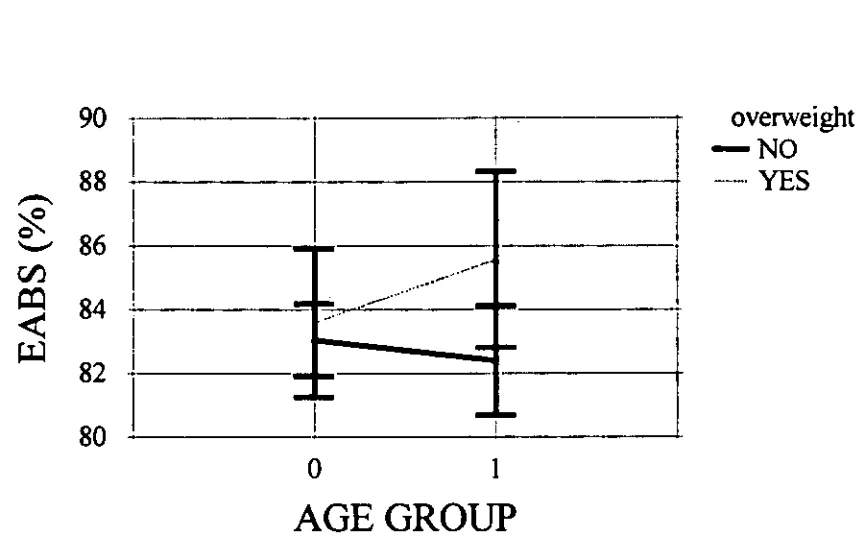


Figure 7.34. Interaction overweight-age group (0= younger, 1= elderly) for energy absorption

There was significant interaction between groups. Figure 7.35 shows that the overweight elderly (102.44 ± 9.83 kNm^{-1}) present a lower maximal stiffness than the young (124.81 ± 23.38 kNm^{-1}) and the non-overweight elderly (108.52 ± 9.83 kNm^{-1}).

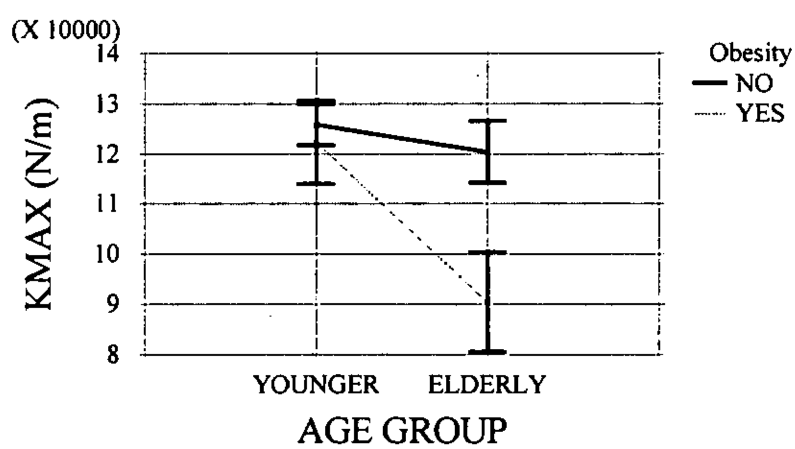


Figure 7.35. Interection overweight-ege group for meximal stiffness

Overweight people presented a lower initial stiffness (26.45 ± 11.72 kNm^{-1}) and final stiffness (67.74 ± 29.26 kNm^{-1}) than non-overweight (40.76 ± 15.17 kNm^{-1} and 91.10 ± 25.87 kNm^{-1}, respectively) (Figs 7.36 and 7.37). Overweight people showed a longer delay between peak force and peak displacement (Tvisco) (0.0041 ± 0.0013 vs. 0.0036 ± 0.0013) (Figure 7.38).

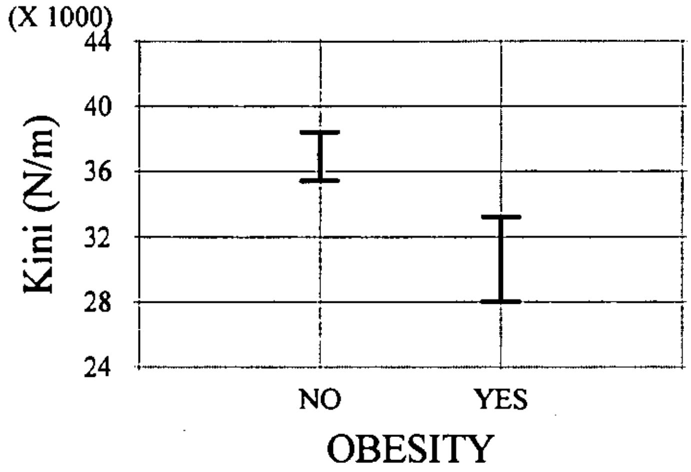


Figure 7.36. Initial stiffness for elderly non-overweight and overweight

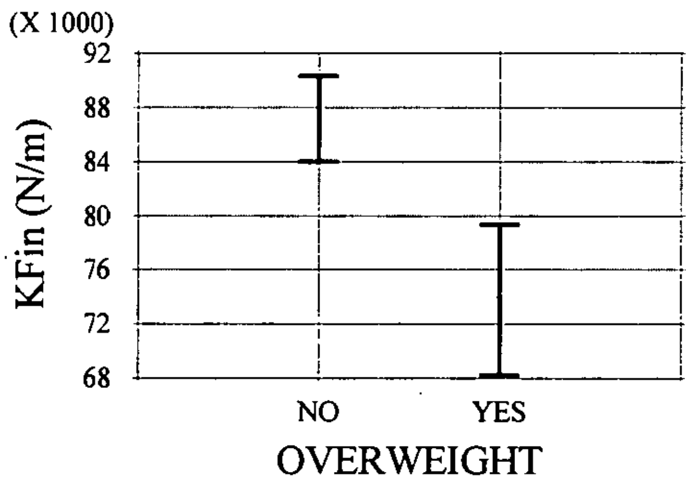


Figure 7.37. Final stiffness for elderly non-overweight and overweight

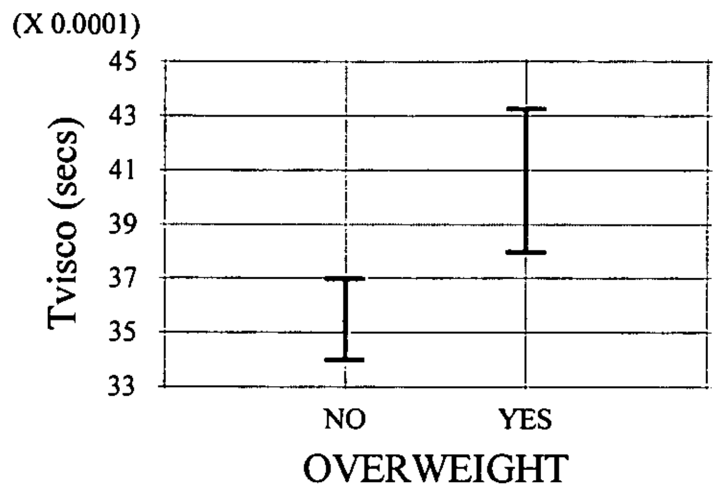


Figure 7.38.Tvisco for non-overweight and overweight elderly

7.10.2.3. Influence of gender in heel pad properties

Time-to-peak force was longer for female (0.0148 ± 0.0025 s) than males (0.0127 ± 0.0028 s) (Figure 7.39). Male participants showed greater energy absorption (p = 0.054) (84.25 ± 7.14%) than female (82.31 ± 6.96%) (Figure 7.40). Time delay between peak force and peak displacement (Tvisco) was also greater in male (0.0043 ± 0.0024 s) than in female (0.0033 ± .0015 s) (Figure 7.41).

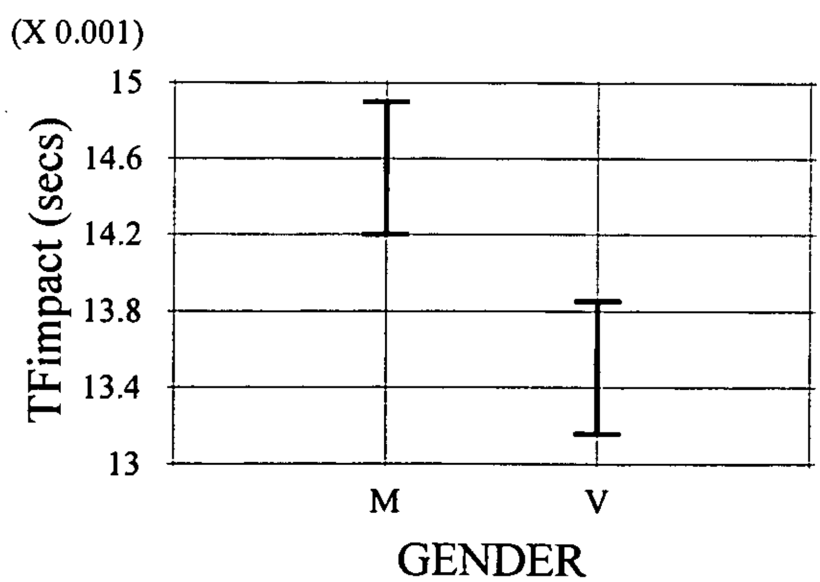


Figure 7.39. Time-to-peak force for femela (M) and male (V)

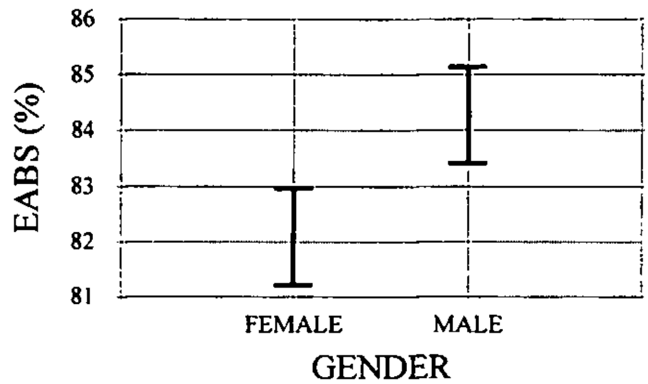


Figure 7.40. Energy absorption for famale (M) and male (V)

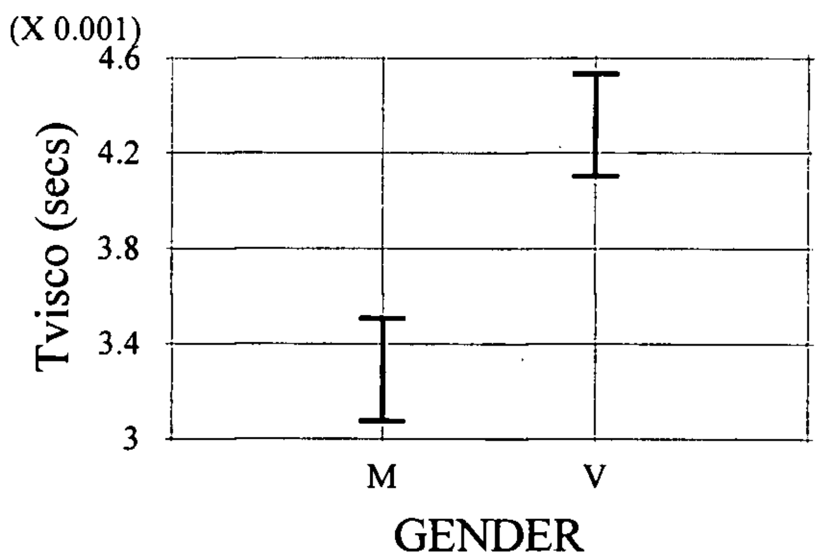


Figura 7.41. Shift tima for femala (M) and mala (V)

Maximal rigidity (126.63 ± 25.034 kNm^{-1}) and Initial stiffness (34.44 ± 14.11 kNm^{-1}, $p = 0.06$)) were greater for males than for females (114.40 ± 2.89 kNm^{-1} and 40.10 ± 16.61 kNm^{-1} respectively) (Figs 7.42 and 7.43).

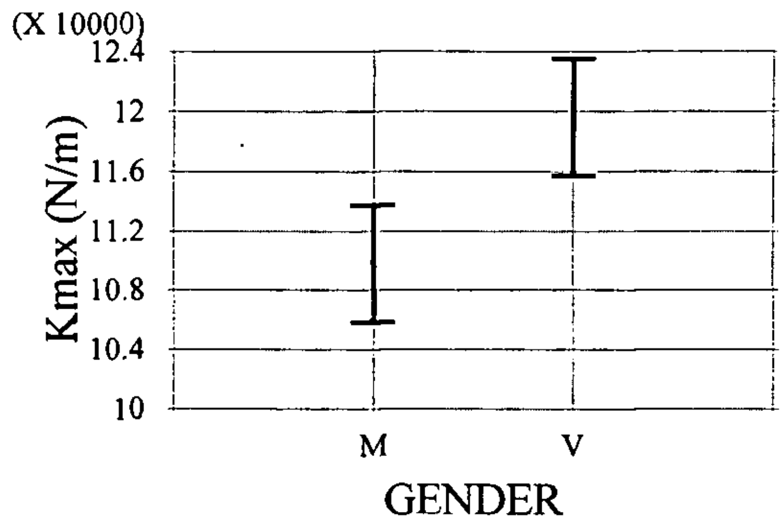


Figure 7.42. Maximal stiffness for female (M) and male (V)

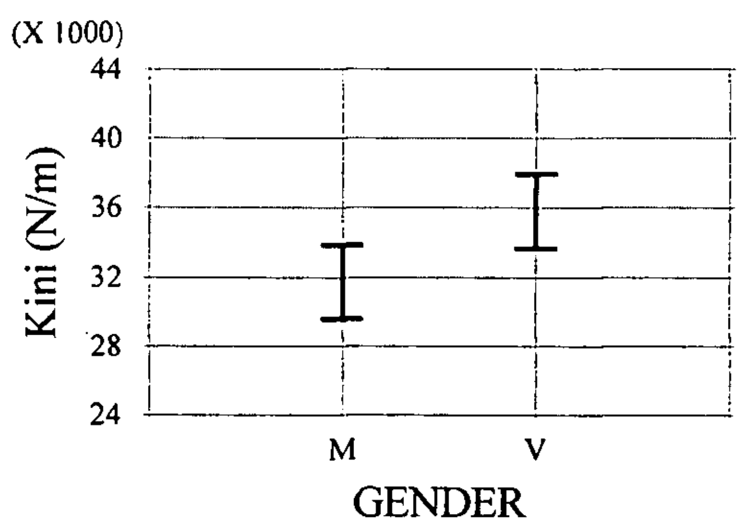


Figure 7.43. Initial stiffness for female (M) and male (V)

7.10.3. Influence of individual characteristics in heel pad mechanics

Differences in elasticity of the heel pad were found due to age and obesity. ANOVA analysis showed that the elastic component in the elderly was statistically significantly lower for overweight (-0.58) than for non-overweight (0.20) (Figure 7.44) which could indicate either a stiffer heel pad in the non-overweight or a more flexible heel pad in the overweight. It was also observed that the younger presented lower score in initial deformation component (-0.160) than the elderly (0.597) (Figure 7.45). No significant interaction was found.

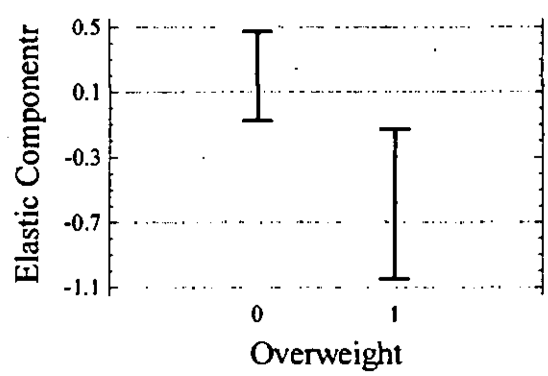


Figure 7.44. Elastic component for overweight (1) and non-overweight (0) elderly

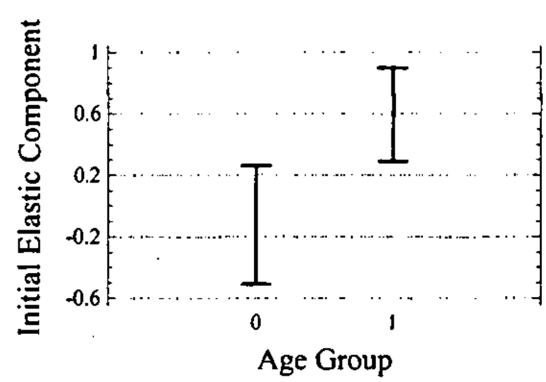


Figure 7.45. Initial deformation component for young (0) end elderly (1).

Relevant differences were found in impact mechanics between the young and the elderly. The young showed high positive significant correlation of impact forces with elastic component (0.81) in the sense that greater forces were related to stiffer heel pad and low negative correlation with viscoelastic component (-0.43) while time-to-peak force showed positive high correlation with initial deformation component (0.77), low negative with viscoelastic component (-0.45) and low negative with elastic component (-0.363) (Table 7.7). Results for the elderly were similar in terms of heel pad mechanics showing high positive correlation of peak force with elastic component (0.905) and low negative with viscoelastic component (-0.463), but also a moderate negative correlation was found with initial deformation component (-0.565). Time-to-peak force showed high negative correlation with elastic component (-0.864) and moderate positive with initial deformation component (0.662). The results of power analysis (Appendix 7. A1) indicated that when a statistically significant correlation was not found, if any, it should be lower than 0.47 for the young and 0.66 for the elderly (Table 7.A1). So, initial elasticity in elderly showed a role in managing impact forces that was not observed in the younger (-0.565 significant in the elderly vs. 0.470 critical value in the young) whilst the role of elasticity in modifying time-to-peak force in the elderly was much greater than in the younger (-0.864 vs. -0.363 respectively, both significant).

Results of correlation analysis for each group separately showed some interesting differences (Table 7.6). Power of test showed that critical value for non-significant correlation would be lower than 0.67 for young male and female and 0.94 for elderly male and elderly female. Young male showed a high positive correlation of impact forces with elastic component (0.81) and moderate negative with viscoelastic component (-0.60). Time-to-peak was highly and positively correlated with initial deformation component (0.89) and low and negatively correlated with viscoelastic component (-0.50). Young women showed similar results, but time-to-peak showed a high negative correlation with elastic component (-0.82) that was not observed in male and, according to power of test; if any, it would be lower than -0.66.

Table 7.6. Correlation results for each study group and for young and elderly

		Initial deformation component	Elastic Component	Viscoelastic component
Fimpact	Young	< 0.471 *	0.805	-0.429
	Young Male	< 0.667 *	0.811	-0.600
	Young Female	< 0.667 *	0.882	< 0.667 *
	Elderly	-0.565	0.905	< 0.667 *
	Elderly Male	-0.679	0.935	< 0.942 *
	Elderly female	< 0.942 *	0.914	-0.696
TFimpact	Young	0.768	-0.363	-0.454
	Young Male	0.890	< 0.667 *	-0.497
	Young Female	0.705	-0.823	< 0.667 *
	Elderly	0.662	-0.864	< 0.667 *
	Elderly Male	0.723	-0.920	< 0.942 *
	Elderly female	0.729	-0.780	< 0.942 *

* indicates that a non significant (n.s.) correlation has been found and that in any case, according to the power of test of the experiment, the value shown would be the smallest significant correlation likely to be found.

The elderly male (Table 7.6) showed similar results, but statistically significant moderate negative correlation was also found for impact forces and initial stiffness (-0.68). Time-to-peak force correlated highly and negatively with the elastic component (-0.92) and positively with the initial deformation component (0.72). Elderly female showed a negative correlation of impact forces with viscoelastic component (-0.70) higher than that in young male (-0.50). (Table 7.6).

The overweight elderly showed high positive significant correlation of impact forces and elastic component (0.93) (Table 7.7). Time-to-peak force correlated negatively with elastic component (-0.82) and positively with initial deformation component (0.76) (Table 7.7). Non-overweight elderly showed similar mechanics, high positive correlation of impact forces with elastic component (0.86) and high negative correlation of elastic component with time-to-peak force (-0.93) However, no correlation between time-to-peak and initial deformation component was observed as in overweight. In any case, the power of test was very low and testing more people would be required for comparison.

Table 7.7. Correlation results for non and overweight elderly.

		Initial deformation component	Elastic Component	Viscoelastic component
Fimpact	Overweight	< 0.894 *	0.933	< 0.894 *
	Non-overweight	< 1.000 *	0.864	< 1.000 *
TFimpact	Overweight	0.760	-0.824	< 0.894 *
	Non-overweight	< 1.000 *	-0.930	< 1.000 *

** indicates that a non significant (n.s.) correlation has been faund and that in any case, according to the power of test of the experiment, the value shown would be the smallest significant correlation likely to be found.*

7.11. Discussion

The heel pad is part of human interface with the ground, thus the understanding of heel pad mechanics under impact loading and its human dependence is very important in the exploration of two of the statements of interest in this research, namely the influence of footwear in heel pad properties and accommodation depending on natural shock absorption. The literature reflects a great inter-individual variability in heel pad properties (Aerts et al., 1995; Cavanagh et al., 1984; Jorgensen et al., 1989b; Kinoshita et al., 1993a, b; 1996a, b), which could be related to the influence that human's characteristics such as age, gender and others appear to have in heel pad properties (Jorgensen et al., 1989b; Kinoshita et al., 1993a, b; 1996a, b; Prichasuck et al., 1994a, 1994b). A great variety of methods and parameters are found in the literature, causing uncertainty about the actual heel pad behaviour under impact loading.

The present study brings a deeper insight of heel pad mechanics in walking loading conditions. It also contributes with a deeper insight of the role of different individual characteristics on heel pad properties. This study provides new and detailed information on heel pad properties for walking loading and on the influence of several individuals' characteristics. On the other hand, the relationship between different parameters is established, allowing for comparison with the literature studies and simplification of future work.

Principal components analysis of heel pad properties with Varimax rotation method identified three mechanical components which could be used for describing the shock absorbing properties of the heel pad. Subsequent analysis showed the influence of age, gender and obesity in heel pad mechanics.

An understanding of viscoelastic materials as they describe the mechanical behaviour of the heel pad as a combination of elastic and viscous elements would be of great help for shoe designers, physiotherapists and others. However, the heel pad is more than a material. It presents a complex structure whose properties can be described by different parameters, probably related to similar aspects of heel pad mechanics. Thus, it is reasonable to suppose that a first step in the mechanical analysis of the heel pad should be the identification of Principal components of mechanical properties of the heel pad which describe its behaviour under impact loading. Factorial Analysis of Principal components identified a little number of components to describe the heel pad mechanics. The method of rotation chosen for factors extraction (VARIMAX) allowed a better understanding of the meaning of the Principal components identified which makes possible to establish a better relationship between the theory of viscoelastic materials and the nature of the heel pad. This is very important for a better understanding of heel pad mechanics and passive interaction.

Peak force (388.1 N to 714.28 N) and time-to-peak force (9.4 ms to 19.6 ms) as well as force-displacement curves obtained in this study were in the range of studies found in the literature (Aerts et al., 1995; Cavanagh et al., 1984; Forner et al., 1995; Kinoshita et al., 1993a b; 1996a, b; Valiant, 1984). *In-vitro* studies in the literature, peak displacement from 9 mm (Valiant, 1984) to 11.3 mm were reported (Cavanagh et al., 1984; Kinoshita et al., 1993a, b; 1996a, b) and this is similar to those found in this study (5.6 to 11 mm). Energy absorption in this study ranged from 67.2% to 94%, which is similar to the range of results in the literature (84 to 99% -Cavanagh et al., 1984; Valiant, 1984- and 75-89%, -Kinoshita et al., 1993a, b, 1996a, b-).

Impact velocity used in the present work was of 0.64 ms^{-1}, simulating walking conditions. Studies found in the literature, however, usually reported heel pad under running or fast walking conditions (from 0.8 ms^{-1} to 1.24 ms^{-1}) (Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1993a, b; 1996a, b; Valiant, 1984); the slowest velocities found in the literature are between 0.57 and 0.74 ms^{-1} (Kinoshita et al., 1993a, b 1996a, b). This makes it difficult to compare results since some workers have demonstrated that the heel pad properties depend on impact velocity (Valiant, 1984). However, impacting mass-velocity combinations resulting in the same input energy have been observed to yield similar results (Cavanagh et al., 1984; Kinoshita et al., 1993a, b). In this study, a mass of 8.155 kg and an impact velocity of 0.68 ms^{-1} were used, so comparison is possible with previous work using a similar combination, for example Kinoshita's (1993a, b; 1996a, b), used 0.57 ms^{-1} and 0.74 ms^{-1} impact velocities and 5 kg mass were used, although using a drop test method could also introduce differences.

7.11.1. Heel pad mechanics

As expected from walking studies of shock absorption (Forner et al., 1995), lower impact forces in the heel pad were related to longer time to peak force. At the same time, lower impact forces were related to greater heel maximal displacement reached in a longer time.

Force and displacement curves obtained in this study are consistent with those showed by different authors (Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1993a, b; 1996a, b). Analysis of force-displacement curve showed non linear and viscoelastic features of heel pad under impact loading common to the literature findings (Cavanagh et al., 1984; Valiant, 1984). An initial region of low stiffness (12.2 to 68.1 kNm^{-1}) was observed, followed by a high stiffness area with mean value of 117 $\pm$ 25 kNm^{-1}; then, the heel pad continued to deform in a medium stiffness area (mean value of 83.08 $\pm$ 28.5 kNm^{-1}) up to peak force. Displacement continued after peak force has been reached. In this context, a positive time delay between peak displacement and peak force was observed (3.6 ms $\pm$ 1.6 ms), which has been related to the viscoelastic nature of the heel pad (Valiant, 1984). After unloading, a residual deformation of about 5 mm was observed at zero force level (Cavanagh et al., 1984; Valiant, 1984). Results for maximal stiffness are consistent with those reported in the literature (Cavanagh et al (1984) reported 130 kNm^{-1} and Valiant (1984) 105.65 kNm^{-1}). However, the values for initial stiffness found in the literature are in the lower limit of the range of values in this study (Cavanagh et al (1984) reported 19 kNm^{-1}, and Valiant (1984) 7.9 kNm^{-1}). Great inter- variability between individual is observed for this parameter and could account for differences in results. On the other hand, differences in the area of contact between the heel pad and the pendulum are likely to affect this parameter (Kinoshita, 1996b). The contact area in this study was 38.48 cm^2 whilst Cavanagh (1984) and Valiant (1984) used a contact area of 60.82 cm^2 and Kinoshita et al. (1993a, b; 1996a, b) one of 12.58 cm^2.

Principal components Analysis results suggest that the heel pad properties under impact loading could be described as the combination of three different components, namely an elastic component acting at initial loading, an elastic component, and a viscoelastic component.

The elastic component included stiffness and heel pad displacement parameters. This component increases as stiffness increases and decreases as time-to-peak displacement and impact duration increase, which could be regarded as an elastic mechanism of force reduction by changing time taken for momentum transfer.

The initial deformation component described a high relationship between initial stiffness, shift time from initial to maximal stiffness and maximal displacement. In such a way that maximal displacement increased as initial stiffness decreased and shift time increased. This component is important as it could reflect an important mechanism of the heel pad for shock absorption by increasing peak displacement and increasing time-to-peak force by mechanical behaviour at initial loading. In a physical sense, organising a disordered bundle of molecules requires energy consumption. Considering thermodynamics, natural state of systems is disorder (maximal entropy), thus organising a disordered system requires energy consumption. Some rubbers have been described as energy absorbers as the disordered bundle of molecules absorb energy to reorganisation under load. From this point of view, it would be possible to define structured materials as those that under load absorb energy by reorganisation of certain internal structure. The heel pad presents a complex structure of fibres and fat (Jahss et al., 1992a, 1992b; Jorgensen & Bojsen-Moeller, 1989a) which could be to some extent disordered at rest. In such a way that under compression some reorganisation would take place absorbing energy. This result is consistent with Valiant's findings (1984) also using a ballistic pendulum related the low initial stiffness to initial reshaping of heel pad in medial-lateral and posterior directions. This reshaping was interpreted as a mechanism of energy absorption. Average displacements observed in lateral and medial directions were of 2 and 2.1 mm, respectively. This medial-lateral displacement occurred during the first 4 or 5 mm (around half of peak displacement) of forward movement of the pendulum in contact with the heel pad. On the other hand, shift time is related to the duration of heel displacement at low stiffness during the initial loading phase, indicating the influence of heel pad reshaping, in such a way that the longer this time the greater the peak displacement. In the present study, this phase lasted for 2.75 ms (range 0.5 - 8.7 ms) in the younger subjects. Low initial stiffness could also be due to joint displacement during impact, but as observed by Cavanagh et al. (1984b) using high-speed film, 3 mm displacement was measured in the ankle after heel displacement so its influence was not observed in the force-displacement curve.

The analysis of this mechanism needs further study since heel pad confinement, which has been described as an effective mechanism of improving heel pad function (Jorgensen et al., 1988), could be related to the initial deformation component. Jorgensen et al. (1988) carried out walking and human drop tests (participant hit a force plate from a given height). He described an improvement of the shock absorbing ability of the heel pad by confinement. However, contradictory results regarding heel pad confinement are found in the literature. Valiant (1984) found a negative influence of firm heel pad lateral compression causing energy absorption reduction (1.1%), peak force increase (28.9 N), peak displacement decrease (10%) and initial stiffness increase which was explained as a possible influence of restraining mechanism on the blood flow which seemed to affect the heel pad shock absorbing properties. In any case, in that research a firm compression was laterally applied to the heel, which actually restricted initial deformation mechanism by avoiding reshaping. However, heel confinement by a shoe heel counter will produce a less firm compression which probably allow some sideways and antero-posterior heel pad movement for initial reshaping. Aerts & De Clercq (1993b), in pendulum tests with humans wearing sports shoes under running conditions, found that force displacement curves of heel pad shod did not present an initial low stiffness region as with barefoot. This was related to the tight lacing of shoes on the foot and thus restricting the sideways movement of the heel pad. In fact the initial stiffness for all shoe-foot systems was 100 kNm^{-1} which is 10 times greater than barefoot values and similar to the results obtained for isolated shoes.

Some authors (Aerts & De Clercq, 1993b) have explained that the good results found by Jorgensen et al. (1988) were due to the effect of heel pad confinement that minimised heel pad bottoming out (Figure 7.46). So, the mechanical overload due to heel pad bottoming out would be registered as high impact forces in tests on non-confined heel pads whilst confinement will avoid overloading, reducing impact forces and improving thus the heel pad properties. But, these results could also be due to the fact that in walking and human drop tests the participant's foot can move in the shoe, probably sideways and antero-posterior movement. That movement could cause the shoe heel counter to be less precisely fitted than in those tests where the participant cannot move it.

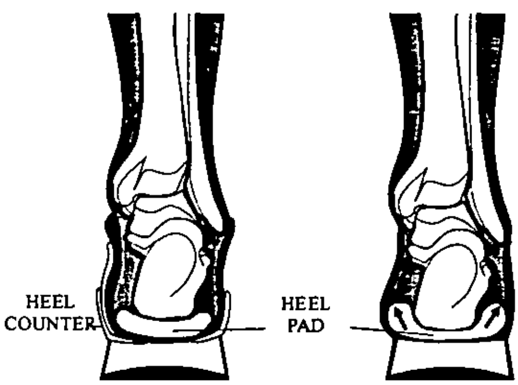


Figure 7.46. Bottoming out of heel pad and heel counter effect.

When the foot is able to move, the heel pad could then be considered as a structured system. The Initial deformation component would be related to mechanical behaviour at initial loading depending on reorganisation. These results could be extended to any material used for shock absorption showing a structure under loading. So, it could be assumed that a heel counter will improve the heel pad function by improving elastic behaviour under initial loading provided that it allows enough reshaping so that initial deformation component is not restricted. This probably happened in some work found in the literature works (Jorgensen et al., 1988; Valiant, 1984), and should be considered in footwear, insole and ortheses design.

Energy absorption and the time delay between peak force and peak displacement defined the viscoelastic component; both are related to the viscoelastic nature of the heel pad.

Elastic and initial deformation components accounted for most of variance in heel pad properties whilst viscoelasticity explained a much lower variance. This could indicate that viscoelasticity is mainly related to properties which are more homogenous within, for example, the subject' heel pad fat, whilst both elastic components would be more related to heel pad structural factors, which seemed to be more sensitive to individual variations. These results suggested that differences between subjects are mainly due to changes in the elastic behaviour. However, this test was done for a given force-time input i.e. frequency content and thus differences in viscoelastic component could arise when varying the frequency content.

Variability in impact force was mainly explained by the elastic component (75.69%, r = 0.85) with a low contribution of the viscoelastic component (17.64%, r = -0.42). Variance of time-to-peak was mainly explained by the initial deformation component (61.0%, r = 0.78). This correlation would suggest that this component contributes to shock attenuation by goveming peak time, consequently modifying momentum transfer by greater deformation.

If the heel pad is considered as a viscoelastic system, its behaviour under impact could be described as an elastic system acting under initial loading which would modify peak time, followed by the combined effect of of a viscoelastic component which would reduce impact forces by absorbing energy, and an elastic mechanism which govern impact force and peak time.

7.11.2. Age influence

Results of this study contrast with the literature findings. Kinoshita et al. (1995) reported higher peak forces in the elderly, but for faster velocity impacts (0.94 ms^{-1}). Jorgensen et al. (1989b), reported that maximal displacement decreased with age, but they used a very slow constant velocity test, which did not simulate the real conditions and would not be reliable for testing a viscoelastic material such as the heel pad. Kinoshita et al. (1996b) reported a lower displacement for fast velocity impact (0.94 ms^{-1}) drop test and lower energy absorption for fast and slow velocity impact (0.57 ms^{-1}). In the drop test by Kinoshita et al. (1993a, b; 1996a, b) the participant was placed lying over the stomach with the foot plant facing upwards. In this position gravity could affect heel pad reshaping, considered an an important mechanism for shock absorption (Valiant, 1984), during initial loading thus reducing maximal displacement and causing changing stiffness measurements. This could explain the 10% difference usually found between drop tests and pendulum results (Aerts et al., 1995; Cavanagh et al., 1984; Kinoshita et al., 1993a, b; 1996a; Valiant, 1984).

The literature results indicate elasticity decrease with age. Prichasuck et al. (1994a, 1994b), related elasticity loss to age based on compressibility index measurements. The compressibility index is a static measurement and viscoelastic materials usually stiffen under dynamic loading. This may be explained as follows: greater viscoelastic properties were measured for overweight and the great proportion of overweight are elderly, thus the viscoelastic component of the heel pad could be considered to play a more relevant role in the elderly than in younger people. No differences were observed in impact forces between elderly and young. Elderly experienced longer time-to-peak force and to peak displacement, together with higher peak displacement and a lower initial and final stiffness, as well as longer shift time (Tcodo).

Changes in the mechanical properties of the elderly heel pad are believed to be caused by a gradual loss of collagen, a decrease in the elastic fibrous tissue and a reduction of water content (Jorgensen, 1990; Prichasuck et al., 1994a, 1994b) and by degenerative biological property changes of the unique system of fibrous structures and fat in the heel pad due to ageing (Jorgensen et al., 1989b). Jahss et al (1992a) in a hystological comparative study, found that there was quantitatively less fat was in the heels of the elderly. The only difference found in the present study was that the elderly had a greater initial deformation component in than the young. This may be explained by degenerative changes that affect the heel pad structure and, therefore, its reshaping (Valiant, 1984) and heel pad properties at initial loading. No differences in the elastic component were found which could be due to obesity. As observed and discussed below, overweight people showed a lower elastic component and therefore a less stiff heel pad. Again it is the elderly that are more usually overweight and obesity may compensate the effect of fat loss and structural degeneration thus maintaining its elastic properties.

In this sense, correlation analysis for young and elderly showed some interesting differences. Moderate negative significant correlation between impact force and initial deformation component as well as higher correlation between time-to-peak and elastic component were observed in the elderly, which could indicate a change in the role of the different components with age. Initial elasticity would gain relevance in modifying impact forces with age and this should be taken into account when considering heel pad properties when designing footwear for the elderly.

7.11.3. Obesity influence

The effects of obesity have been obtained only for the elderly and further research is needed before drawing any conclusions about the younger obese.

According to Kinoshita et al. (1996b), the lower the fat content, the lower the energy absorption. However, no differences in energy absorption between overweight and non-overweight were observed in this study. Overweight people showed lower impact forces and longer time-to-peak than both the non-overweight and the young. Peak displacement and time-to-peak were also greater than for non-overweight and young. Energy absorption and time delays were greater in overweight.

Maximal stiffness, initial stiffness and final stiffness were lower for the overweight than for non-overweight. The elastic component was greater (stiffer heel pad, i.e. lower elasticity) for non-overweight elderly than for the overweight. Obesity is understood to increase fat content in the heel pad leading to increased pressure in a closed space, which should result in a loss of the elasticity of the heel pad and greater (stiffer) elastic component, as shown by the greater compressibility index in the overweight humans (Prichasuck et al., 1994a). However, those results do not apply to the elderly although increasing stiffness and loss of fat have been observed with age. Obesity could affect the elderly in different ways to the younger person by compensating for fat loss due to age and structure degeneration thus maintaining elasticity in the overweight elderly. This could explain the results from this study.

This is very important for rehabilitation and the design of footwear for the elderly since the elderly are more frequently overweight. At the same time, the representation of the heel pad as the combination of three Principal components should feature in further studies for a better understanding of the influence of obesity on the elastic and viscoelastic components.

Nevertheless, as with the age effect analysis, both the elderly and the younger were tested under the same impact conditions as the purpose of this work was a comparison between these age categories The effect of actual peak forces of overweight people during walking was not considered, which could affect the absolute level of forces and heel pad properties. In consequence, studies of the influence of obesity in ground reaction forces (GRF) would be necessary to extrapolate these results to footwear design.

7.11.4. Gender influence

The results of this study showed that the females had shorter time-to-peak force with a lower energy absorption and time delay as well as lower maximal and initial stiffness. This is in agreement with Jorgensen (1989b) who found a more rigid heel pad, as well as greater peak displacement in males. This could indicate a less viscoelastic heel pad in females.

No global differences were found between the young male and female. However young males did not show significant negative correlation (if any, it should be lower than 0.66) of time-to-peak force with the elastic component unlike young females. This could be due to structural differences in heel pad between young males and young females.

Gender differences have been related to body structure and/or hormonal differences. A thicker heel pad has been found in men which could be due to greater growth hormone concentration; whereas in women, the high estrogen level might influence the stiffness of the fibrous structure (Jorgensen et al., 1989b; Prichasuck et al., 1994a; 1994b). Differences in impact forces associated with elderly women were partially explained by the viscoelastic component. This could be related to hormonal changes occurring after the menopause affecting the properties of the heel pad fat.

Passive interaction

On the other hand, initial results from this research (see Chapter 6) suggeste that accommodation seems to depend on the properties of the insert material beneath the human heel since the same shoe was worn in all cases. Forehead transmission appeared to be inversely related to the material's dynamic stiffness, which would control impact perception. This could be reasonably extended to global stiffness between the ground and the mechanoreceptors, including the heel pad properties. Considering results of this research on heel pad mechanics, stiffness at initial loading which is related to structure and heel pad confinement is important for shock absorption. Mechanical components similar to those found in the heel pad exist in any material with an internal structure that re-orientates under load. In this sense, it may be possible to describe a passive interaction between foot and shoe that would modify underfoot stiffness and change impact perception by a combination of heel pad and material properties. Heel pad confinement and shod testing of its properties show results that suggest some interaction. The interface between underfoot material and heel pad is likely to modify initial stiffness as well as global stiffness of both. The heel will sink deeper in a softer material thus increasing confinement since a deeper heel cup produced. In this sense, De Clercq et al. (1990) defined a mechanical coupling between heel pad and shoe affecting behaviour under initial loading in such a way that mainly the shoe responded at the initial impact. At the same time, the mechanical coupling of heel pad with insole underfoot material will change the material's structure and its initial stiffness thus modifying the behaviour at initial loading of both heel pad and insole material. The heel pad bottoming out and confinement and the subsequent. However it is also important to know how perception is affected by this passive interaction without movement. In this area further research is needed for a better understanding of the role of active and passive interaction in impact perception.

That passive interaction may play an important role in impact mechanics is supported by the results reported by Lafortune et al. (1995) using human pendulum testing to investigate the role of knee flexion in controlling impact with different EVA interfaces. That work provided support for the dominant contribution of heel fat pad and material interface in the lower leg stiffness and cushioning during initial impact loading. Wall force peak decreased as knee angle increased while tibia shock and load frequency increased with knee angle. Peak increased, time-to-peak decreased, transient rate and mean power frequency increased from soft EVA to hard EVA to barefoot. Hard and soft foams produced substantial reductions of effective axial stiffness of the leg (EASL) compared to barefoot and this stiffness was significantly lower for the soft EVA than the hard EVA (58.15 kNm^{-1} to 168.87 kNm^{-1}). There was high correlation of effective axial stiffness (EASL) and wall peak force transient rate, while moderate correlations were observed with other variables such as peak force, peak shock and shock transient rate. EVA interface demonstrated far greater cushioning and shock attenuation ability than initial knee angle. Large differences found between results from two densities of EVA supports that confounding effects during *in-vitro* evaluation of footwear cushioning properties are due to kinematic adaptation. EASL was completely defined by the interface conditions. The combined stiffness of heel pad and material interface would thus govern completely the rate of application of force during initial impact.

7.12 Conclusions

The results of this experiment supported the contention that shock absorption in the lower leg could be adequately described as the result of combination of three different mechanical components, namely an elastic component in series elastic and viscoelastc components combined in parallel. The elastic system acts first at initial loading then the elastic – viscoelastic system the response of which reflects rates of loading.

The elastic and initial deformation components performed a more important role in impact mechanics than the viscoelastic component. The elastic component accounted, in general, for most of differences in impact forces, while initial deformation component was correlated to the peak time. The viscoelastic component showed a low correlation with the impact force and time-to-peak.

Differences in heel pad mechanics due to age, gender and obesity were observed mainly in relation to initial deformation and elastic components. Both seemed to be related to structural human-dependent elements of the heel pad and explained most of the variance in heel pad properties.

Interesting differences were observed in the elderly. The initial deformation component was greater than in the young and showed correlation with impact forces whilst the elastic component explained part of the variability of time-to-peak. It is suggested that the initial deformation component is related to reorganisation of the heel pad by means of reshaping in medial-lateral and posterior directions, which could be affected by age degeneration and heel pad confinement. This should be considered in footwear and orthesis design.

It was suggested that a greater elastic component (stiffer heel pad) for the non-overweight elderly than for the overweight is related to compensation of age elasticity loss by increased fat content associated to obesity. Overweight elderly showed lower impact forces and longer time-to-peak than the non-overweight and the young. Peak displacement and time-to-peak were also greater for the overweight than for the non-overweight and the young. Energy absorption and time delays were greater in the overweight. Maximal stiffness, initial stiffness and final stiffness were lower for the overweight than for non-overweight. So, overweight could compensate age-related loss of shock absorbing ability of the heel pad.

Female participants presented a shorter time-to-peak force together with a lower energy absorption and time delay and a lower maximal and initial stiffness than did male participants. The elastic component correlated to time-to-peak in young females but not in young males, probably due to structural differences in heel pad.

These results should be considered in footwear and orthosis design for maintaining and improving the shock absorbing properties of the heel pad, mainly by considering elastic and initial deformation components for adequate heel counter or heel cup design.

The structure and materials of the human heel pad is responsible for their good mechanical properties. Further research should focus on relationship between these properties to explain the origin of individual differences. The development of models of the heel pad comprising the three components described in this Chapter will be of great value for footwear designers. The simplest mechanical model representing the heel pad as in one of the most frequently models used for modelling viscoelastic materials that include a dashpot and a spring combined in parallel (viscoelastic and global elastic components respectively) in series with another spring (initial stiffness) (Figure 7.47). This could be a very good general representation of the heel pad and further research should be devoted to it. Modification of initial stiffness by footwear, insoles design or age, as shown in this study would mainly affect the series spring constant whilst humans differences would probably be related to both springs. Further research of this aspect when shod will enable study of the insole effect.

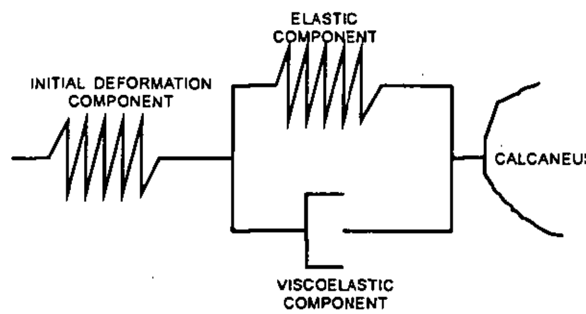


Figure 7.47. Heel pad model proposed from the results of Principal Component Analysis

On the other hand, the relationship between accommodation and insert stiffness described in Chapter 6 was redefined considering the mechanical coupling of heel pad and insole. It was suggested accommodation depends on underfoot stiffness as a result of the mechanical coupling between insole material and heel pad.

This report, so far, has been shown that machine test simulating walking forces is unable to reliably predict the behaviour of insole materials during walking due to human accommodation which appeared to be related to the material's stiffness. Further studies on the properties of materials were deemed necessary. On the other hand, a method for heel pad testing was developed which allowed to explore the complex impact mechanics of this important natural shock absorbing system showing that the properties of the heel pad depend on individual characteristics. As a result, it was also suggested that accommodation is related to human stiffness. In consequence, passive interaction was defined as the mechanical coupling between humans and materials. These results and methods define more clearly the nature of further studies needed to explore the remaining research fields including material and human testing as well as pendulum testing of shod people.

Appendix 7A. Power of test

Power of test is the probability of detecting a difference among groups by means of a statistical test when such a difference exists. Power analysis depends on the statistical technique used. In this work *Analysis of Variance* and *Correlation Analysis* were used. For Analysis of Variance, power of test depends on number of participants and trials, as well as on number of levels of each component. It was important in this research so to know whether statistically significant differences not found by Analysis of Variance are small enough to have no relevance at a practical level, or whether more participants or trials are needed to obtain valid results. In correlation analysis, power depends on number of participants and when comparing results among groups, the smaller the statistically significant correlation likely to be found at a given probability level has to be known to yield relevant conclusions about differences between groups.
A study of **power of test was done for all the correlation analysis**. This study enabled an evaluation of the probability of finding statistically significant correlations of a given size for each analysis. Due to the very conservative way of computing test power, an 80% probability of establishing and effect of a given size is usually considered as reasonable.
Table 7.A1 shows the effect size for an 80%-90% probability of establishing significant (Alpha = 0.05) correlation for each analysis. If no significant correlation between two variables is found in a given analysis, then any possible effect should be lower than values shown in Table 7.A1 which represent the greatest correlation likely to be found by increasing n.

Table 7.A1. Critical values for p=0.05 with power of test for 80-90% probability for each population group.

GROUP	Size	Minimum Significant Correlation which could be expected to be found
All Data	54	0.385
Young	36	0.471
Elderly	18	0.667
Young male and young female	18	0.667
Elderly male and elderly female	9	0.942
Overweight elderly	10	0.894
Non overweight elderly	8	1.000

Table 7.A2 shows the effect size for an 80% probability of establishing significant (alpha = 0.05) results for all the variables. The asterisk shows variables for the significant effects found in each study. One asterisk indicates significant differences between male and female. Two asterisks indicate statistically significant differences

between young and elderly while three asterisks indicate statistically significant differences between overweight and non overweight elderly. If no significant effect is found for a given variable, then any possible effect should be lower than the values shown in Table 7.A2.

Teble 7.A2. Power of test for ANOVA Analysis. (*, significent differences for gender; **, ege and ***, overweight)

	Age and Gender		Overweight and gender in elderly	
Variable	Magnitude	% of signal	Magnitude	% of signal
Dmax (m)	$2.3\ 10^{-4}$ (**)	3.15	$2.47\ 10^{-4}$ (***)	2.88
Eabs (%)	1.24 (*)	1.50	1.18	1.43
FIMP (N)	13.32	2.39	14.43 (***)	2.71
Kfinal (kN/m)	5.20 (**)	5.79	4.64 (***)	6.00
Kini (kN/m)	2.43 (*,**)	6.40	2.18 (***)	7.88
Kmax (kN/m)	4.57 (*)	3.83	5.07 (***)	4.80
Kmean (N/m)	3.91	4.71	4.10	6.02
Tcodo (s)	$3.2\ 10^{-4}$ (**)	11.62	$2.7\ 10^{-4}$	6.73
Tdmax (s)	$4.7\ 10^{-5}$ (**)	2.85	$5.3\ 10^{-4}$(**)	2.84
Tdura (s)	$8.1\ 10^{-4}$	2.01	$9.6\ 10^{-4}$	2.29
TFimpact (s)	$4.0\ 10^{-4}$ (*,**)	3.03	$4.1\ 10^{-4}$ (***)	2.66
Tvisco (s)	$2.5\ 10^{-4}$ (*)	7.70	$2.3\ 10^{-4}$ (***)	6.99

8

Selection of materials

8. SELECTION OF MATERIALS

8. Selection of materials

Two main considerations of concern in the literature survey were:

i. The experiments reported used either hardness or density as criteria for choosing the range of experimental materials. However, it was not clearly demonstrated that these properties are relevant to shock absorption so the range of materials used may have been inappropriate. Other mechanical properties such as loss tangent, compressive stiffness, etc. should also have been criteria for selecting materials.

ii. According to the review of accommodation, plantar pressure distribution may play an important role in accommodation onset in comfort and in impact perception. Robbins & Gouw (1991) described how plantar pressure tolerance is not the same for all the areas of foot plant and if plantar pressure exceeds the threshold of certain areas, an avoiding behaviour will transfer pressures to other more tolerant areas. At the same time, the literature survey revealed that both pressure beneath the heel and plantar pressure distribution were related to impact perception and to comfort. Thus, according to accommodation hypothesis the plantar pressure could be the type of loading that related better to shock absorption than impact forces and the mechanical properties of materials relevant to plantar pressure distribution may help to predict materials performance in use. For this, a machine testing method for simulating plantar pressure would be needed. No method was found in the literature so such a method had to be developed as part of this research.

A total of 52 commercially available insole materials were selected. This range of materials was tested by different machine methods: static compression, IBV method simulating impact forces and the new method (developed as part of this research) to simulate plantar pressures. Hardness and density were also measured. Its ability to predict shock absorption, pressure distribution, rearfoot movement and comfort during walking is described in later Chapters

This new machine method was used to test a suitable range of materials. Statistical techniques were used to reduce the mechanical properties obtained from these methods to a smaller set of properties to select the study materials.

8.1. DEVELOPMENT OF A MACHINE TESTING METHOD SIMULATING PLANTAR PRESSURES

8.1.1. Introduction

To conduct this research, a new test method was developed to study the behaviour of materials during walking. This new test was designed to study the mechanical properties of materials under that simulated plantar pressure loading during conditions associated with walking. Materials in the study were viscoelastic and non linear whose behaviour is dependent on frequency and magnitude of load and for this reason the method developed was based on simulating pressure-time history under different areas of the foot. This included defining the loading history for testing and the identification and selection of parameters to be deduced from the tests.

The work was developed in three stages:

i. Definition of the testing method and definition of load magnitude and duration, equipment test protocol. To define the loading, plantar pressures were registered during walking.
ii. Identification and selection of study parameters. This work required signal processing and development of software.
iii. Experimental set up for testing a sample of existing materials and consequence reduction of study parameters and loading sets.

8.1.2. Definition of test method

Peak pressures and times under the foot depend on the plantar location. A general pattern depending on footwear, population, velocity and certain other factors. In essence, greatest pressure is under the heel, followed by the hallux and then the first, central and fifth metatarsal heads and finally midfoot plant and the rest of toes. A great range of peak pressures and times are to be found in the literature depending on the plantar location. For this reason, a single loading condition for material testing could give rise to confusing and unrealistic results. On the other hand, as already stated, impact perception is related to heel pressures and according to the Robbins' theory also is related to metatarsal heads pressure. So both have to be considered.

For these reasons, load history under the different areas should be simulated. A set of loading conditions representing plantar pressures and times will give a better understanding of the behaviour of materials under the different foot plant areas. In consequence, the first experiment conducted was to measure the plantar pressure distribution during walking.

8.1.2.1. Definition of loading history

Using the IBV methodology, pressure was measured under eleven different foot areas. The laboratory equipment (Figure 8.1) included a set of instrumented insoles, photocell barriers, digital chronometer and a PC for data acquisition.

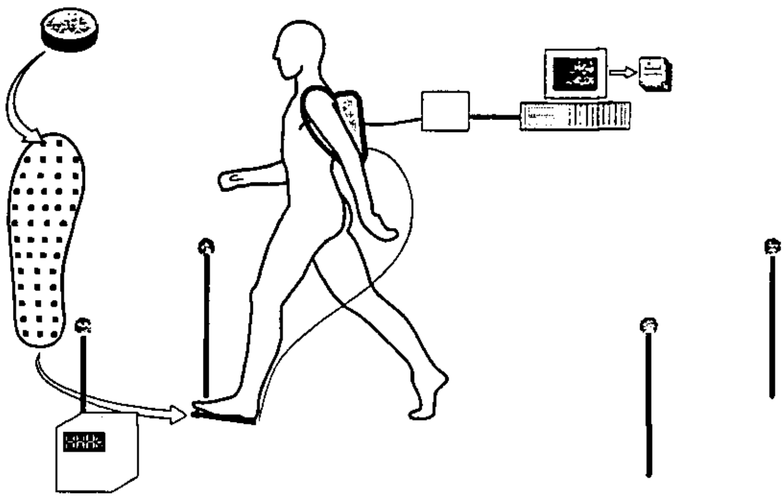


Figure 8.1. Laboratory set up for plantar pressure distribution testing.

The pressure distribution was registered with an instrumented insole [the IBV Biofoot®] placed inside the shoe. It consists of a flexible insole with up to 64 sensors with a total sampling rate of 600 Hz. The sensors are piezolectric ceramics that return an electric signal when pressure is applied. The ceramics have been distributed according to foot physiology in such a way that a greater density of sensors is placed under bony areas (beneath the toes and metatarsal heads) to measure local variations in pressure. The insoles together with electronics register pressure at each sensor at a high speed and generate a substantial amount of data. For analysis, the foot plant was divided into eleven physiological zones (Figure 8.2). Pressure for each area was computed as the average of the pressure data for all the sensors in this area, the maximum pressure of this average value was considered for analysis as well as the maximum pressure of a single sensor in the area. To compute the central tendency of pressure for each area the median value was used.

The photocell barriers drove the digital chronometer. The distance between the barriers was fixed and the time was controlled so that only trials within ± 10% of a fixed velocity for each participant (free speed) were accepted.

TE Laterel heal

TC Cantral Hael

TI Internel hael

PME Lataral midfoot plant

PMC Central midfoot plant

PMI Internal midfoot plant

C5M Fifth metatarsal head

MM Central metatarsal heads

C1M First metatersal haad

RD Toes plant

PrD Hallux plant

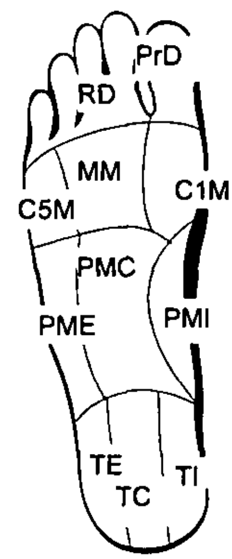


Figure 8.2. Areas into which the foot plant is divided.

Ten volunteers (5 male and 5 female) took part in the experiments, walking on 9 different casual shoes (a representative range of shoes on the market). Maximum absolute pressure under each area, the maximum mean from all the sensors in a given area and the time of pressure loading were computed for further analysis.

Time of pressure peaks, contact time for each area and other variables were first considered. But, a correlation analysis from previous experiments showed that the variance of all variables was clearly a function of the maximum mean pressure of each area. As a result, only this variable was considered for this study.

Regarding the time-to-peak pressure, statistical distributions showed two clearly differentiated peaks. The maximum values below 800 ms which corresponded to the first maximum were considered to represent the functional aspect of pressures whilst the other values were considered to be residual values.

The results obtained for each area and gender are shown in the Table 8.1.

Table 8.1. Pressure and contact time under each foot plant area.

	Pressure (kPa)		Time (ms)	
	male	female	male	female
First toe	200	500	400	1000
Rest of toes	90	100	300	1000
1st metatarsal head	300	500	600	800
Central metatarsal heads	400	400	600	600
5th metatarsal head	200	200	600	800
Medial arch	37	90	100	400
Lateral arch	100	100	400	400
Metatarsal bump	50	100	600	800
Medial heel	200	400	400	600
Lateral heel	400	500	200	400
Central heel	500	300	400	600

The results were reorganised in increasing order of pressure and time as shown in Table 8.2.

Tabla 8.2. Pressure and time combinetions

	Plant area	Pressure (kPa)	Time (ms)
Male	Internal arch	37	100
Male	Central midfoot	50	600
Male	Toes	90	300
Female	Internal arch	90	400
female	Central midfoot	100	800
Female	External midfoot	100	400
Femele	External midfoot	100	400
Female	Toes	100	1000
Female	Fifth metatarsal head	200	800
Male	Internal heel	200	400
Male	fifth metatarsal head	200	600
Male	Hallux	200	400
Female	Central heel	300	600
Male	First metatarsal head	300	600
Male	External heel	400	200
Male	Central metatarsal head	400	600
Female	Internal heel	400	600
Female	Central metatarsal head	400	600
Female	External heel	500	400
Female	First metatarsal head	500	800
Male	Central heel	500	400
Female	First toe	500	1000

As expected, a great variety of times and pressures were found in the different locations, the greatest pressures under the central heel, first metatarsal head and first toe. It is not realistic to test such a great number of combinations, but a smaller representative number of combinations may be identified. Pressure was reduced to five magnitudes at 100 kPa intervals. Times show more variance but, in general, a fast and a slow time can be identified in each pressure interval.

Thus the pressure magnitudes and times identified as representative were as follows

- Pressure kPa: **100, 200, 300, 400 and 500**
- Time ms: **400** and **800**

As a result, the dynamic cushioning test consisted of five different pressure magnitudes applied for 400 and 800 milliseconds giving 10 different loading sets.

Each test consisted in applying five compression cycles with a time lapse of 1300 ms between two consecutive peaks (Figure 8.3). This time corresponds to the time between two consecutive steps during walking.

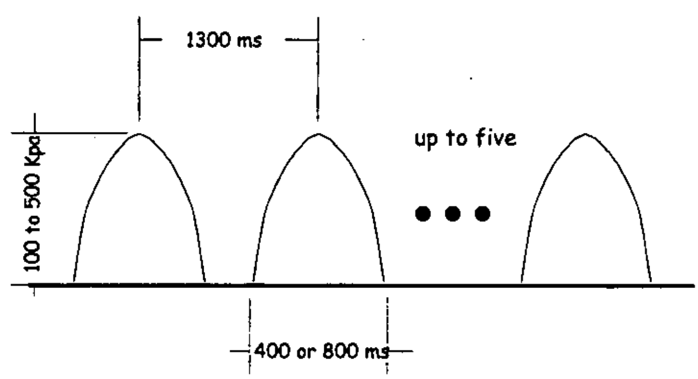


Figure 8.3. Loading history

8.1.2.2. Equipment for Testing and measurement

This test was carried out using an Universal dynamic INSTRON 8501 testing machine, with a maximum applicable load of 100 kN. However, for this method a load cell of 5 kN was used. The test was automatic by controlling the displacement of the actuator. The force and displacement signals were registered by a portable computer connected to the machine by means of a PCMCIA data card in the serial port.

8.1.2.3. Testing protocol

A cylindrical clamp of diameter 50 mm was used to compress the plate or insole of material placed in the lower compression plate of the machine. Rectangular plates of material were tested instead of circular samples to avoid border effects.

Each material is tested under ten different conditions. Loads of 20, 40, 60, 80 and 100 kg were applied each for 400 and 800 milliseconds duration. A preload of 25% of maximum force was applied. Thus, the peak forces applied were 15, 30, 45, 60 and 75 kg respectively.

Force and displacement were registered in volts through the control console of the testing machine by the personal computer. The conversion factors to be applied for force and displacement were as follows

$$F(kg)=50.97 * V \qquad F(N)=499.5 \times V$$
$$D(mm)=7.59 * V \qquad D(m)=7.59 \times 10^{-3} * V$$

where V is the signal in volts.

The data was registered in a file for further processing in MATLAB. Matlab format was used for storing the data to save space and operating time.

8.1.3. Identification and selection of parameters

8.1.3.1. Signal processing

Specific software was developed in MATLAB 4.0 under Windows for processing the data. This program was called ***cusi.m*** and is listed in the appendix A4. In this programme, the force data is converted to pressure ($P = F/S$, where S is the clamp surface: 0.0025 m^2). The program restricts the signal to the peaks only.

Some conditioning of the signal was required before processing. Filtering to eliminate noise was made first. Because force and displacement were acquired through the Data Card of a Personal Computer there was a time delay between both signals. The data card registers a datum each [1/sampling frequency] seconds for all the channels in a sequential manner. Thus, it takes a datum each channel simultaneously and simultaneously. So, force is registered at instant N and then displacement after a 1/(2*sampling frequency) delay. Interpolation techniques were used to eliminate this delay. Functions available in Matlab library for interpolation by B-Splines were used for this purpose (see appendix A4).

Once force and displacement were in the same time base, a series of functions developed to compute the identified parameters were used. Some of the Matlab functions previously developed for pendulum testing were implemented in this software to calculate mechanical properties.

8.1.3.2. Identification of parameters

From force and displacement, as well as force-displacement curves (Figure 8.4) a series of parameters were identified to describe the mechanical behaviour of materials many of which were similar to those developed for heel pad analysis (Chapter 7). These can be divided into *impact* and *material* parameters, which included force-displacement, and pure parameters.

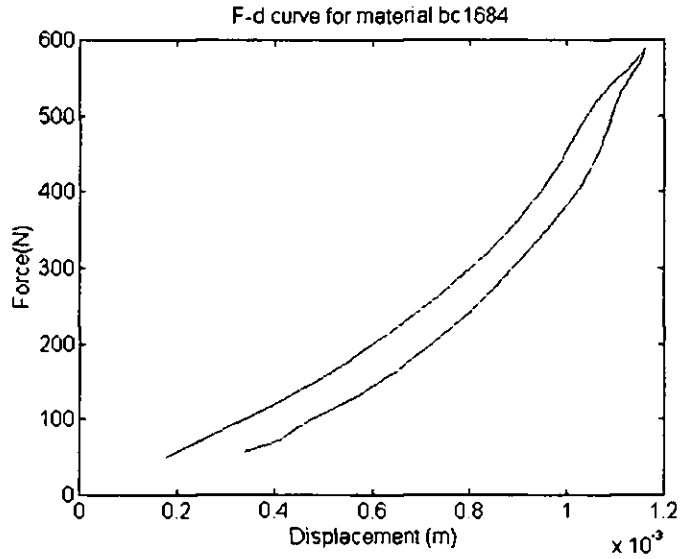


Figure 8.4. Typical force-displacement from dynamic cushioning test

Impact parameters included peak force and time to peak.

The *pure parameters* included:

.- Peak displacement (Dmax) and time (TDmax).

.- Integration of displacement on time (Area) which was devised as a "displacement impulse" to some extent expected to be related to recovery ability of materials and so to material bottoming out.

.- Residual displacement when loading force reaches the zero value (Rebote) which directly measures recovery of materials.

The *force-displacement* parameters included:

.- Energy absorption (Eabs) obtained as the area of the hystheresis loop by integration of force against displacement.

.- Loading work (Wdefor) (integration of force against displacement during loading phase) which represents the energy necessary to compress the material.

.- Stiffness parameters similar to those obtained from pendulum test (Chapter 7): initial (Kini), maximum (Kmax), final (Kfinal) and stiffness at peak force value (Kfmax).

.- Time of shift from initial to maximal stiffness (Tcodo) as a measurement of the duration of initial low stiffness region.

.- Time delay between peak force and displacement (Tvisco) as described in Chapter 7.

All were obtained for each loading set. These parameters were selected because they relate to specific mechanical properties or phenomena as follows:

Property/Phenomena	Parameters
Elastic properties of materials	Stiffness, time of shift, peak displacement and time-to-peak.
Bottoming out	Residual displacement, loading work and displacement integral.
Viscoelastic properties of materials	Energy absorption and time of delay

8.1.4. Method set up

Fourteen parameters were obtained for each material from each loading case, which made a total of 140. It is reasonable to believe that these parameters were correlated (i.e. different parameters describe the same information regarding materials' behaviour and variance) for each testing condition as well within different testing conditions. Thus, the number of parameters and also the number of testing conditions could be reduced for subsequent analysis. This informs the process of materials selection for the experimental study. For this a hierarchical clustering analysis of variables was done using the absolute Pearson correlation coefficient as distance measure to reduce the number of parameters.

8.2. Selection of materials

To obtain a reasonable range of insole materials' properties, a large number of the available insole materials were assessed.

In the first phase, a total of 52 materials were studied, from which a manageably smaller number of them were selected for the next phase of the experiment. Plates from new and existing materials were prepared for testing by manufacturers collaborating in this project.

To aid the selection of materials, properties obtained from static compression testing, the IBV impact test and dynamic cushioning test developed in the course of this research, as well as hardness Shore A and density were obtained for each material. A statistical analysis of variables reduced the number of variables to study, then a number of materials was selected to cover the range of values of the properties chosen.

8.2.1. Sample of materials

Materials selected at this stage reflected the variety of materials most commonly used for insoles in footwear. Materials tested in this phase were combinations of three basic constituents:

Rubber, either natural or synthetic. Rubber needs a process called vulcanisation (heat transformation) to modify its elasticity.

EVA (Ethyl-Vinyl-Acetate). This is a polymer widely used in footwear for both insoles and soles and is available in different densities and compositions

Polyethylene. This polymer is available in different densities from low to high depending on the complexity of the molecular structure; in this project medium density PE was used.

Polyurethane is also often used for insoles but due to technical problems at the time of producing samples it was not possible to include this material in the analysis.

Table 8.3. Materials used in the project

Material	Composition	Differences between materials in the group
AB-1, AB-2, AB-3	Cellular (foam) rubber (mix of rubber, inorganic charge, plastificants and additivas as helpers and for vulcanisation and swelling)	Three different formulas with the same charge. The only difference is the chemical make up of the rubber mixture
BB-1, BB-2	Cellular (foam) rubber (mix of rubber, inorganic charge, plastificants and additives as helpers and for vulcanisation and swelling)	Two different formulas with the same charge. The only difference is the chemical make up of the rubber mixture. Both charge and rubber mix-up are different from the former group.
BC-1, BC-3	Cellular (foam) rubber (mix of rubber, inorganic charge, plastificants and additives as helpers and for vulcanisation and swelling)	Two different formulas with the same charge. The only difference is the chemical make up of the rubber mixture. Both charge and rubber mix-up are different from the former groups.
BT-1, BT-2	Cellular (foam) rubber (mix of rubber, inorganic charge, plastificants end additives as helpers and for vulcanisation and swelling)	In this group rubber mixtures are different form the other groups and within them.
ESP-I, ESP-II	Doreco® (mix of copolymers of EVA with a mixtures of rubbers, charges and additives)	Variations in polymers and chemical charges
TWN, PJ, TERMO	Doreco® (mix of copolymers of EVA with a mixtures of rubbers, charges and additives)	Variations in polymers and chemical charges
G-2	Doreco® (mix of copolymers of EVA with a mixturas of rubbers, charges and additives)	Variations in polymers and chemical charges
6505-K, 7506-K, 7503-K, 5003-K	Expanded Polyethylene	Different hardness and densities

Vulcanisation time is considered important for rubber-based materials AB, BB, BC and BT. Groups ESP, TWN, PJ and TERMO were manufactured in two different thicknesses. G2 is a special mixture of Eva and rubber created for this project.
Eight different families of materials, differing in basic composition and additives (tables 8.3 and 8.4), were tested. In each group, different materials were included considering the criteria described in Table 8.4.

Table 8.4. Materiels in each group

Material	Criteria
AB-1, AB-2, AB-3	Vulcanisation time. AB16 means the material AB1 with 6 minutes of vulcanisation and AB18 with 8 vulcanisation minutes.
BB-1, BB-2	Vulcanisation time. BB29 means material BB2 with 9 minutes of vulcanisation.
BC-1, BC-3	Vulcanisation time. BC38 means the material BC3 after 8 minutes of vulcanisation.
BT-1, BT-2	Vulcanisation time. BT112 means material BT1 with 12 minutes of vulcanisation.
ESP-I, ESP-II	Different thickness. ESPIT1 and ESPIT2 show the same haidness and density but different thickness.
TWN, PJ, TERMO	Different thickness. TWNT1 end TWNT2ESPIT1 show the same hardness and density but different thickness.
G-2	Different thickness. G2T1 and G2T2 show the same hardness and density but different thickness.
6505, 7506, 7503, 5003	Different compositions of expanded polyethylene.

8.2.2. Static compression test

This test was set up according to the standards DIN 53.545, ISO 604 for compression testing of synthetic materials. The scope of these standards was to analyse the behaviour of synthetic materials under uniaxial compression loading to determine resistance and shape change. Care was taken to ensure that the uniaxial compression was achieved.
This test consists of applying a ramp of controlled displacement on the material sample at a given constant velocity. To establish the magnitude of pressure to be applied, the plantar pressure distribution during walking was used in setting up the dynamic cushioning test. In this case the 75 percentile of all data was taken as maximum pressure (Table 8.1).
Thus, the test was designed to reach a maximum pressure of 500 kPa, a typical maximal value observed during walking.
The test were conducted using a universal testing machine [SERVOSIS], according to the standard DIN 51223 for static tensile, compression, bending and torsion as well as combined testing.

The material sample was compressed between two polished steel cylindrical flat clamps of 50 mm. diameter with round edges in such a way that the applied force was evenly distributed across the entire sample surface. Compression was achieved by controlling the actuator displacement at a given velocity. In this test the materials were loaded at a velocity of 0.08 mms^{-1}. Preload was fixed at zero force. The test ended when a force of 750 N (corresponding to a pressure of 500 kPa) was reached in approximately 110 seconds. The external integrity of every sample was tested at the end of the test.

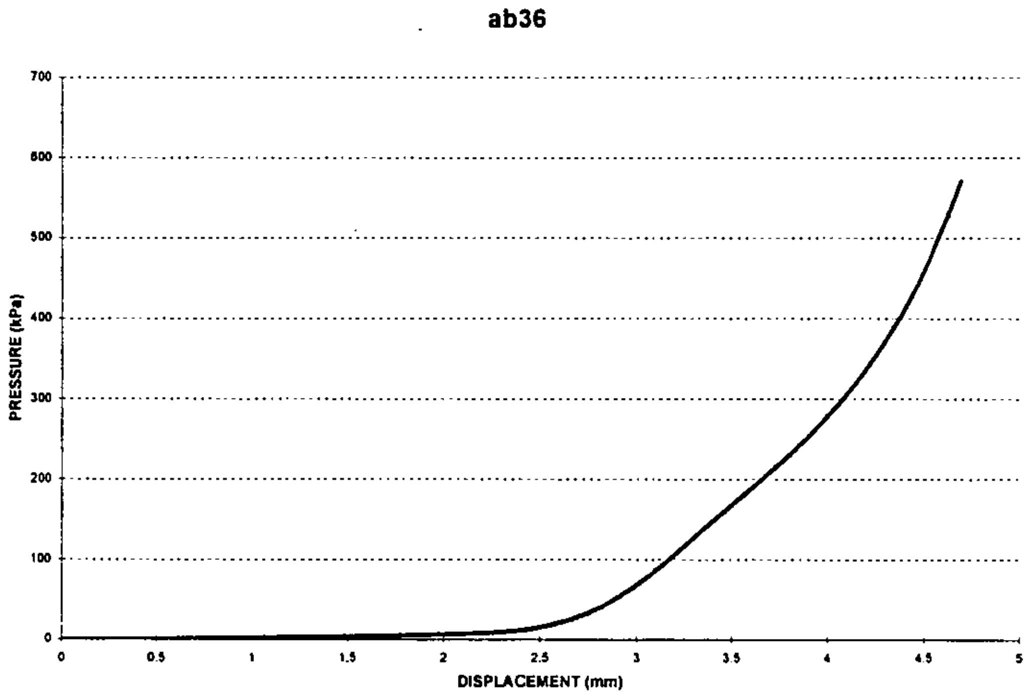


Figure 8.5. Pressure-displacement curve for material AB36

Force and displacement were recorded for further analysis. A special software called *parama* (Appendix A4) was developed under Matlab 4.2 for Windows to compute displacement and stiffness in 11 pressure steps between 0 and 500 kPa as well as the global stiffness (between 0 and 500 kPa). The stiffness between two pressure magnitudes was determined from the slope of the linear regression between force and displacement in the pressure interval. A typical force-displacement curve is showed in the Figure 8.5.

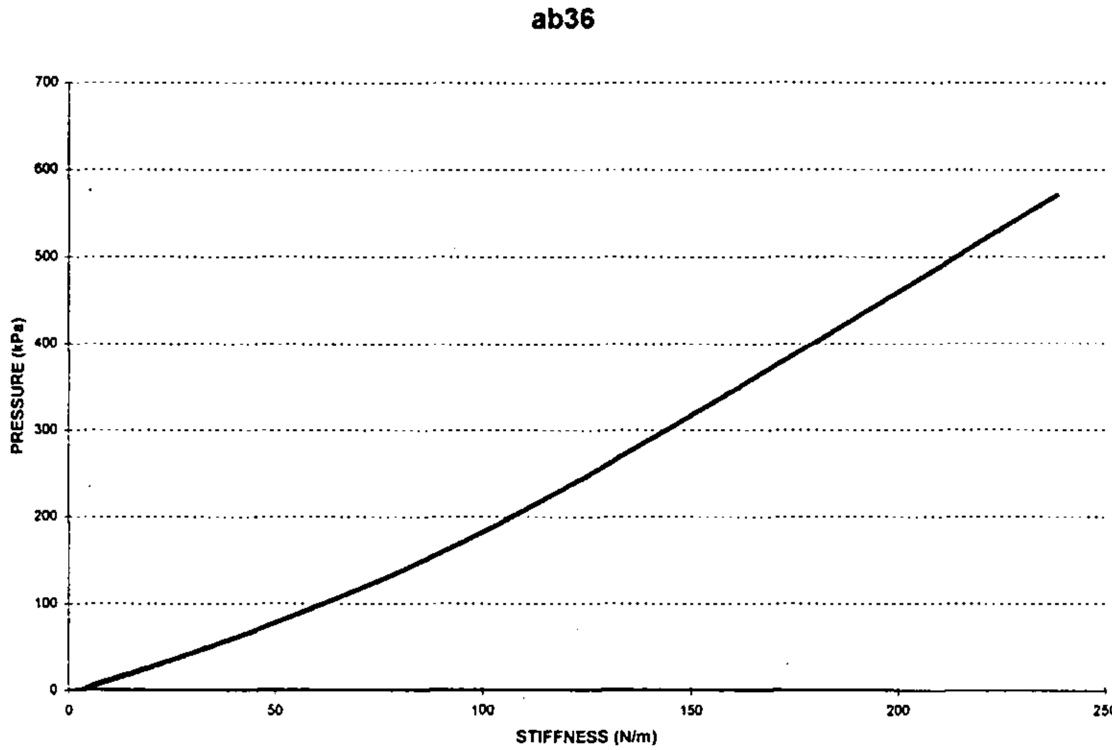


Figure 8.6. Stiffness-pressure curve for material AB36

The force data was converted into pressure units with respect to the sample surface. The initial region of the curve was eliminated because of anomalous data due to initial set-up of materials and testing machine given that preloading was not applied. This usually corresponded to less than 1.5 mm displacement and 50 kPa pressure.

The results obtained from this test were:

- *Material displacement for 50 kPa (D50), 100 kPa (D100), etc. upto 500 kPa (D500) pressure magnitude, as well as the initial displacement under preload (D0).*
- *Material stiffness for 50 kPa (R50), 100 kPa (R100), etc. up to 500 kPa (R500). The total stiffness was also obtained (Rt).*

A typical stiffness curve is showed in Figure 8.6 showing a non-linear behaviour, stiffness increases as pressure increases.

8.2.3. Dynamic testing of impact properties

This test has been described in detail in Chapter 6 (García, 1984). A Universal dynamic Instron 8501 testing machine is used to simulate walking impact forces (Figure 8.7). Force and displacement signals are acquired and the Fast Fourier Transform made by means of a dual channel spectrum analyser. Dynamic stiffness and loss tangent at frequency intervals of 5 Hz between 0 and 40 Hz were obtained as were mean values.

Figure 8.7. INSTRON 8501 testing machine

Eight impacts are applied to each material controlling the displacement of the machine actuator. Plates of materials were tested by a cylinder clamp of 50 mm diameter with round edges. Samples were not cut-out from the plates to avoid border effects.

An impact force of 50 kg was applied with a preloading of 20 kg and a rise time of 20 ms. The results obtained from this test were:

.- *Dynamic stiffness at 5 (Rig5), 10 (Rig10), ... and 35 Hz (Rig35).*

.- *Loss tangent at 5 (Tan5), 10 (Tan10), ... and 35 Hz (Tan35).*

Mean stiffness (Rigidin) and loss tangent (Tandin) for all the frequencies were also calculated.

8.2.4. Reduction of mechanical variables to study

As a result of mechanical testing, a total of 191 parameters were computed from three tests for each material. Density, thickness and hardness were also computed. Statistical techniques for data reduction and classification were used, first, to reduce the number of parameters for further study and, second, to select a reasonable number of materials representing the full range of parameters selected for manufacturing insoles for the next experiments. The statistical analysis was done using SPSS 7.5.2.s for Windows

A graphic and descriptive analysis of the data was done to check their validity and to eliminate anomalous data and outliers.

The statistical procedure used for variable reduction was a hierarchical clustering analysis between variables using the between groups method and the absolute value of Pearson correlation coefficient as distance measurement.

An initial analysis was done to reduce the number of variables needed from the testing methods. A significant correlation $R > 0.8$ was considered (64% of variance explained).

In this sense, from this first approach the variables were reduced to 74 (less than half) as follows:

- **Hardness and density**
- **Static compression (5 parameters out of 22).**
  - *Displacement at 250 and 300 kPa (D250 and D300) pressure magnitude*
  - *Stiffness at 150, 300 kPa and global (R150, R300 and RT)*
- ***Dynamic impact testing (3 out of 16)***
  - *Mean loss tangent **(tandin)***
  - *Mean dynamic stiffness **(rigidin)***
  - *Loss tangent at 5 Hz **(tan5)***
- ***Dynamic cushioning testing (64 out of 140)***

For simplicity, the loading conditions are referred to by two numbers, the first represents the pressure magnitude from 1 to 5 corresponding to 100 to 500 kPa and the second is the duration, 4 means 400 ms and 8, 800 ms.

A high correlation was found between area, wdefor and rebote which are parameters describing a global compressive behaviour of the recovery material. This was not the case of stiffness parameters that, to some extent, describe the shape of the force-displacement curves (i.e. the materials mechanical behaviour)

- *Residual displacemant at 200 kPa pressure magnitude and 800 ms of duration (loading 28) **(Rebote28)***
- *Energy absorption for 100 kPa pressure and 800 ms duration (loading 18) **(Eabs18)***
- *Energy absorption at 24 loading **(Eabs24)***
- *Peak displacement at 18, 28 and 48 loading **(Dmax18, 28, 48)***
- *Initial stiffness for 34, 44, 54, 18, 28, 38, 48 and 58 loading **(Kini18, 28, 34, 44, 48,.54, 58, 38)***
- *Final stiffness for all pressures for 800 ms **(KfinalN8)***
- *Maximum stiffness **(Kmaxnn)**, Time of peak displacement **(Tdmaxnn)**, Delay time between peak force and peak displacement **(tvisconn)** and shift time from initial to maximum stiffness **(tcodonn)** for all conditions*

A summary of parameters reduction is shown in the Table 8.5, blank cells correspond to those variables eliminated from the study.

Teble 8.5. Reduction of peremeters

Loading Set	18	28	38	48	58	14	24	34	44	54
Eabs	■						■			
Rebote		■								
Area										
Wdefor										
KFmax										
Kini	■	■	■	■	■					
Kmax	■	■	■	■	■	■	■	■	■	■
Kfinal	■	■	■	■	■					
Dmax	■	■		■						
TDmax	■	■	■	■	■	■	■	■	■	■
Tvisco	■	■	■	■	■	■	■	■	■	■
Tshift	■	■	■	■	■	■	■	■	■	■

From these results, area, wdefor and Kfmax were eliminated as output from the new testing method.

8.2.5. Materials selection

At this stage, the number of variables was further reduced to a reasonable number for material selection. Ten, or less, variables were considered reasonable for choosing the materials sample. The criteria for selection were as follows:

1. ***Variables obtained from measured signals rather than calculated ones (regression, integration, etc.).***
2. ***Variables with a physical meaning***
3. ***Correlation greater than R=0.5 (moderate, 25% of variance explained). This is rather low, but necessary at this stage, to reduce the number of variables.***

As a result, 9 variables were chosen as follows:

♦ **Static compression (2)**

- *Displacement al 250 kPa (D250)*
- *Stiffness at 300 kPa (R300)*

♦ **Dynamic impact testing (2)**

- *Mean dynamic stiffness (RIGIDIN)*
- *Mean loss langent (TANDIN)*

♦ **Dynamic cushioning testing (3)**

- *Maximum stiffness from 48 loading (Kmax48)*
- *Maximum stiffness from 28 loading (Kmax28).*
- *Delay time between peak displacement and peak force at 58 loading (TVISCO58).*

♦**Density and hardness.** Hardness is a surface property resulting from simple indentation tests and could be used for material testing as an alternative to more complex machine tests.

Ten materials were considered adequate to keep the testing to a reasonable duration and cost. The main objective of the experimental part was to examine the relationship between the different biomechanical variables, subjective and mechanical variables. Thus, the statistical analysis to be done was based on correlation and regression techniques; the materials to be selected should show values for the selected variables that covered the whole range of possible values for each variable. Extreme and intermediate values should be included, for example the hardest, the softest and some intermediate hardness materials should be included. To do so for 52 materials and 9 variables would be rather complicated.

The statistical procedure followed for the selection was:

i. *A clustering analysis (K-means Cluster between groups) was done for all 52 materials using the 9 selection variables to obtain 10 groups. The materials most representative of each group (central in the cluster) were first selected; the number of materials chosen was smaller but still greater than ten.*

ii. Dispersion charts for all nine selected variables were drawn with the smaller sample of materials. From these charts, materials were selected to represent the range of values of the variables.

iii. A further criteria was to include materials for all the chemical families included in the sample (ab, bc, bb, bt, esp,...) (Table 8.3).

The materials finally selected were

- *AB19, AB36, AB37, AB210*
- *BB28, BB210*
- *BT112*
- *BC18*
- *ESP1T1*
- *C7506K1*

All materials were 5 mm thick, except ESP1T1, which was 3 mm thick. Table 8.6 shows thickness, hardness and density of selected materials each with an identification code that will be used in this study.

Teble 8.6. Properties of selected meterials

MATERIAL	Thickness (mm)	Hardness (Shore A)	Density (kg/m³)	Code
ESP1T1	3.08	30.8	.0003	T1
BB28	5.15	11.4	.0006	BB28
AB37	5.87	34.6	.0009	AB37
BC18	5.61	9.6	.0007	BC1
BB210	5.15	9.6	.0006	BB21
AB210	5.43	29.6	.0010	AB2
AB19	5.71	31.4	.0008	AB1
BT112	6.06	18	.0009	BT1
AB36	5.77	31.4	.0010	AB36
7506K1	6.17	24.2	.0001	K1
Mean		23.06	0.0007	
PERCENTILE25		16.32		
PERCENTILE75		29.80		

Some dispersion charts are included to show that the range of properties was covered (Figure 8.8).

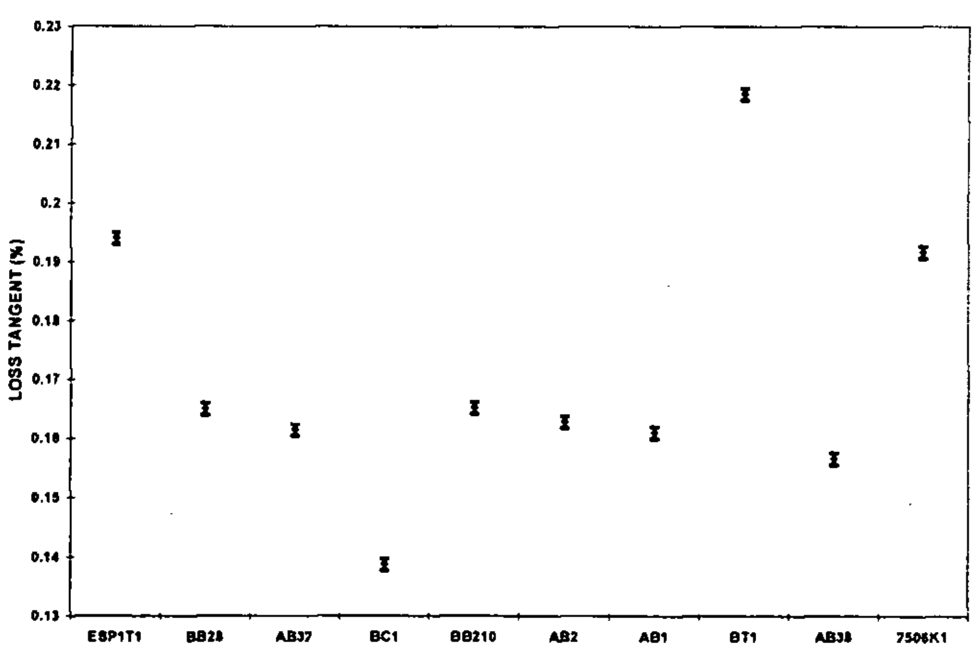


Figura 8.8. Loss tangent for the materials sample

Test samples of insole materials for the experiments were developed by the author and manufactured from the selected materials by a supplier. The same geometry (shape and dimensions) and covering material were used for all test samples. The insoles were tested again using the same mechanical tests to obtain actual properties. Some comments are necessary with respect of insole properties. The effect of the covering material with possible border effect and size introduced some differences between material properties when tested as a plate and when tested as an insole. The most important differences were observed in loss tangent that decreased drastically when tested as insole and this should be considered in a further analysis. No coherent explanation, apart from possible chemical changes in the manufacturing process, was found.

The literature search highlighted the importance of pressure distribution for comfort and impact perception. Results described in Chapter 6 suggested a reverse relationship between stiffness and human accommodation and indicated the need for a different type of machine testing to predict the performance of materials during walking. Different machine tests were conducted to select the materials to make the insoles that were afterwards used for walking and machine testing according to the design of experiments for this research. The new machine testing methods described in this chapter were used to characterise the materials be used as insole materials in subsequent experiments. These experiments are described in the following chapters.

9

HUMAN TESTING

9. HUMAN TESTING

9. Human testing

Experiments involving human testing were of two types:

.- active while humans walked

.- passive using the instrumented pendulum with no humans movement.

Two techniques were used:

- subjective - to gather the opinion of the participants
- objective - to measure various biomechanical parameters.

To describe the study, the experiment is divided into:

.- **Walking experiments,** which include *objective* and *subjective* testing for the biomechanical analysis of people walking while wearing different insoles. The objective tests carried out comprised shock absorption, pressure distribution and rearfoot movement analysis during walking. Subjective testing included the analysis of comfort and impact perception.

.- **Pendulum experiments** were conducted with volunteers standing whilst wearing different insoles into a shoe.

Walking experiments are described in this Chapter. The material and methods used for collecting biomechanical and subjective variables, with relevant results based on some partial objectives are described.

Following the general experimental design, the data collected was further analysed together with pendulum results and material testing to investigate the different statements. However, for a more comprehensive appreciation of the work and results obtained, it is appropriate in each case to describe the methods and materials along with the results, discussion and conclusions arising from first analysis of data.

9.1. Insole materials and footwear

A total of ten insoles were constructed with the selected materials [Chapter 8]. All the insoles were the same thickness, pitch, geometry and covering material (leather). All the insoles were tested into the same shoe — a men's street casual Oxford shoe. It featured leather upper and hard rubber sole and was originally designed to include a removable insole. This shoe was taken as reference to avoid fitting problems when changing the insole.

The performance of the shoe in shock absorption may be due mainly to its influence on walking kinematics, the heel pad confinement or the material itself. The third research field explored argues that the effect of material is much lower than the effect of heel confinement and the shoe alone. So by testing different insole materials in the same shoe, the effect of the shoe in walking kinematics was minimised depended basically on the materials and geometry of the rear part of the shoe. Thus the effect of insole materials only was studied. For heel pad confinement, the effect of the shoe was the same for all insoles. However, and considering previous results of this work, a passive human-insole interaction may occur depending on insole material and heel pad properties giving different degrees of heel pad confinement (i.e. heel pad-insole interfaces).

The characteristics of the shoe from static compression and dynamic impact testing are shown in the Table 9.1. The static stiffness was calculated to be between 30 kg and 50 kg which is the magnitude that impact forces generally reach. Stiffness and loss tangent showed the usual behaviour in terms of frequency, the former increased whereas the latter remained practically constant.

Table 9.1. Properties of the shoe used for testing

Static stiffness (N/m)	*Loss Tangent*	*Dynamic stiffness (Nm^{-1})*
$632.65\ 10^3$	0.169 ± 0.001	$(31.26 \pm 0.11) \times 10^5$

At this point, it was decided not to include shoe properties in the analysis. Theoretically, the properties of the shoe could be added directly to those of the insoles by modelling springs and dashpots in series. The shoe is stiffer than insole materials whilst the loss tangent is similar. Since the prime interest of this research was the analysis and comparison of insole-effect using regression and correlation analysis, the shoe properties need not be considered. Nonetheless, if analysis of the absolute values of impact forces and accelerations were of interest, the effect of shoe materials should be considered.

9.2. Participants

Ten healthy non-obese males were selected for the experiments thus avoiding affects due to gender, obesity and age in impact mechanics (Chapter 7). Participant's characteristics are shown in Table 9.2.

Table 9.2. Participants characteristics

	Mean	Maximum	Minimum	S. D.
BMI	23.01	27.73	19.88	2.41
AGE	22.90	28.00	19.00	2.51
HEIGHT (cm)	177.65	183.00	173.00	2.89
WEIGHT (kg)	72.51	83.00	63.00	6.33

The age range was between 19 and 28 years (mean of 22.9 years). Only one participant had a BMI slightly over 27 (27.73). All statistical analysis was completed using SPSS7.2. and Statgraphics Plus 2.1 for Windows for multifactor ANOVA. In general, the variables are not independent but as the interest of the study was in measured and meaningful parameters rather than in combined parameters Manova was not used.

9.3. OBJECTIVE TESTING

Objective testing included shock absorption, pressure distribution and rearfoot movement analysis whilst subjective testing was a study of comfort and impact perception.

Objective testing included techniques to register biomechanical variables related to shock absorption and proprioception. Participants performed objective testing while wearing the ten insoles in the same type of shoe, as well as barefoot when the technique allowed this.

Different biomechanical tests were conducted to register variables that could be related to impact absorption and attenuation as well as to proprioception. The analysis of results at this level was conducted to reflect on some of the questions raised in the aims of this study. Such reflections are discussed for each technique separately.

9.3.1 Shock absorption

The analysis of shock absorption during walking with different insole materials was mainly to collect data for further comparison with other biomechanical and material variables and impact perception, as this is the major subject of the research. However, as all the insoles were tested into the same type of shoe, differences between them should be due to accommodation, insole material or passive interaction. Some partial goals also directly investigated from this analysis were:

- To check whether impact conditions differed significantly by simply changing insole material. The goal of the proprioceptive system has been described as keeping either impact forces or head acceleration at a safe, constant value. Variations in biomechanical variables were analysed with this respect.
- To study the relationship between biomechanical variables.

9.3.1.1 Material and methods

The methodology used for registering vertical impact forces and the shock transmitted through the human body was as described in Chapter 6. Basically this consisted of simultaneously registering vertical ground reaction forces by means of a force plate (Dynascan®) and acceleration by accelerometers fixed at the shank and the forehead (Figure 9.1).

Ten male participants performed 5 acceptable trials walking across a force plate while wearing each of the 10 different insoles into the same type of shoe. Barefoot is the hardest walking condition and was tested last to avoid influencing accommodation. The insoles were tested in a random order. Impact perception was also collected in this study. This required participants to actually test 12 different conditions since the first was assigned as a reference for impact perception and this was repeated two more times.

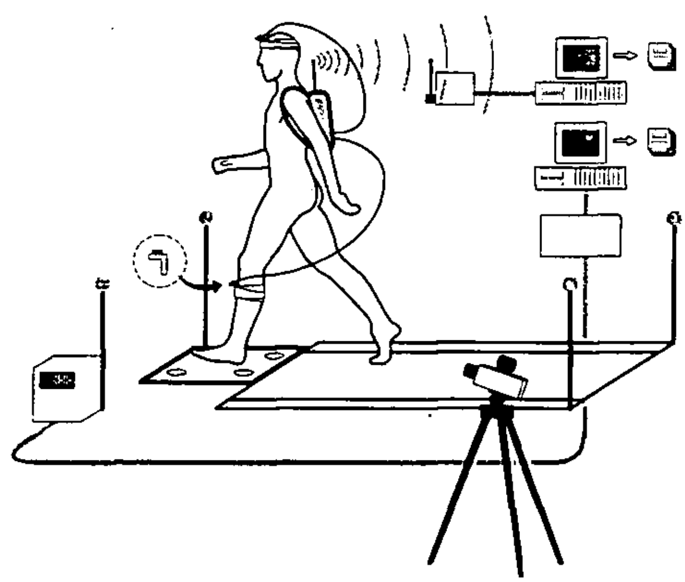


Figure 9.1. Laboratory set-up for shock absorption analysis

The parameters recorded for further analysis have been described in Chapter 6 and included peak of impact force (MF1), shank peak acceleration (Mt) and forehead peak acceleration (Mc). Time-to-peak impact force (Tfi) and rate of loading (Ratac) were also measured. The Ratio between peak acceleration at both locations (MTMf1) [Mt and Mc] and impact force, between peak accelerations (McMt) were also calculated.

Impact force was measured in body weight units (BW), acceleration in "g" units and time in milliseconds. All variables were normalised (nd) values with respect to barefoot results (for instance head acceleration in a participant with an insole divided by the head acceleration when the same participant walked barefoot).

The statistical analysis was done at two different levels: all participant-conditions combinations (ALL level) and mean by insoles (Insole level, which would happen if an insole is worn by an average participant, within the range of the sample that took part in the experiment).

Analysis of variance (ANOVA) considering participant and insole as a factor as well as the second order interaction between them was done to study differences in all the biomechanical variables to assess if any variable remains constant for identification as a possible system goal. Multiple range Bonferroni analysis post hoc was done to assess differences between homogenous groups. Assumptions for normality were tested for each variable using the Lebene test for homogeneity of variance and Kolmogorov-Smirnov (K-S) test for normality.

Factor Analysis of Principal components was done to ascertain the underlying structure of impact parameters and to reduce the dimension of the study for further correlation analysis with other biomechanical variables. Principal component method using Varimax rotation with Kaiser normalisation was done at all levels. Components were chosen with Eigen value greater than 1 when rotated and explaining more than 80% of total variance. Variables score in components was considered when greater than 0.5. A minimum of five to ten cases is required for each variable included in the Factor Analysis. In this sense, at All level 100 cases were available since the aim of the study was the interaction between insole and participants, thus each combination participant-insole was considered as a case. This analysis was not done at insole level because of lack of power of test since at both levels only 10 cases were available. At this level, Pearson's correlation analysis was done to assess relationship between variables.

Statistical analysis of correlation and factor analysis of principal components was done using SPSS7.5.2. for Windows. ANOVA was done in Statgraphics plus 2.1 for Windows.

9.3.1.2. Results

The global descriptive statistics for all variables including the coefficient of variation (COV = standard deviation/ mean) are presented in Table 9.3 as a measure of variables dispersion. In general, forces and accelerations registered were in the range previously reported in the literature search (Chapter 6).

Anova results showed some statistically significant differences at both participant and insole level. **Impact forces** showed differences between participants (Figure 9.2 -

Multiple range test), and between insoles when barefoot data (DES) was eliminated from the study. The observed interaction was statistically insignificant. Values ranged between 0.42 and 0.46 (8.6%) and significant differences found were around 6%. As expected, walking barefoot yielded the higher impact forces.

Table 9.3. Shock absorption results

	Mean	Sd. Deviation	Maximum	Minimum	COV
MF1	0.47	0.12	0.93	0.22	0.266
MF1ND	0.68	0.14	1.00	0.47	0.201
RATAC	$17.0\ 10^3$	$14.05\ 10^3$	$87.54\ 10^3$	$6.60\ 10^3$	0.828
RATACnD	34.98	26.46	137.2360	16.60	0.756
TMF1	0.981	0.367	1.758	0.555	0.374
TMF1nd	0.022	0.014	0.039	-0.110	0.635
MC	1.65	0.25	2.50	1.19	0.149
MCnd	0.81	0.10	1.00	0.62	0.127
MT	2.61	0.87	7.16	1.61	0.334
MTND	0.61	0.18	1.0081	0.38	0.293
MCMT	0.67	0.12	0.94	0.32	0.180
MCMTnd	1.414	0.37	2.26	0.71	0.261
MTGMF1	0.91	0.21	1.59	0.64	0.227
MTMF1	5.70	1.03	10.31	3.55	0.181

Anova results showed some statistically significant differences at both participant and insole level. **Impact forces** showed differences between participants (Figure 9.2 - Multiple range test), and between insoles when barefoot (DES) was eliminated from the study. The observed interaction was statistically insignificant. Values ranged between 0.42 and 0.46 (8.6%) and significant differences found were around 6%. As expected, walking barefoot yielded the higher impact forces.

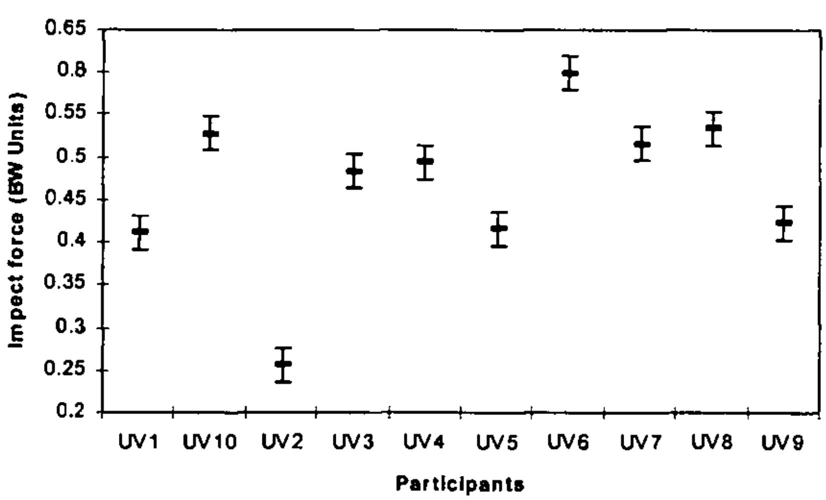


Figure 9.2. Impact forces at participant level

The impact force normalised with respect to barefoot also showed differences between participants and some differences between insoles (Figure 9.3), BB21 resulted in lower forces than AB1 and BT1. This variable represents differences in humans' behaviour with respect to barefoot depending on the insole material.

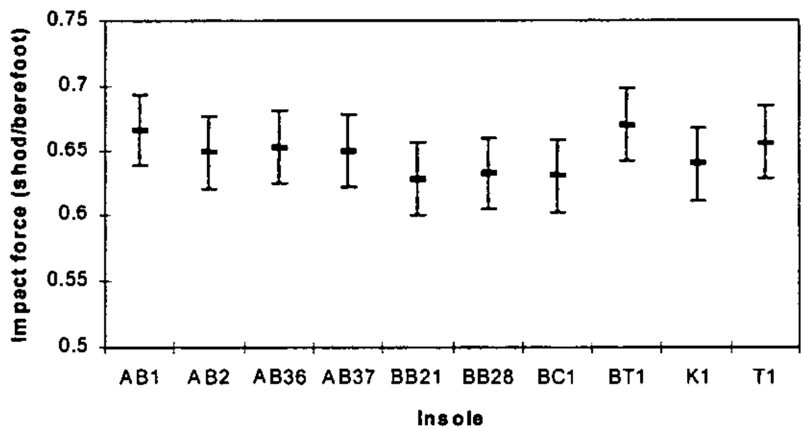


Figure 9.3. Impact forces with respect to berefoot at insole level

Time to peak force showed differences only between insoles whilst **Rate of impact force loading,** both absolute values and normalised with respect to barefoot, was only higher barefoot with no differences between insoles. Differences were however found between participants. In this sense if perception as seen in the literature (LaFortune et al., 1995b; Milani et al., 1995; 1997) is related to this variable, the present result suggest that it could be considered as a system goal.

There were statistically significant differences in accelerations. Tibia acceleration, absolute values and with respect to barefoot, presented differences for both participants and insoles (Figure 9.4). Barefoot were higher. At the same time, several homogenous groups of insoles were observed as resulted from Multiple Range test. Greater differences were found between humans.

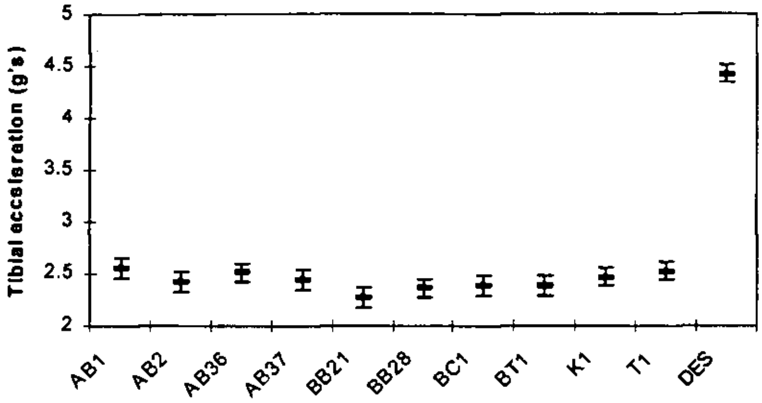


Figure 9.4. Tibia acceleration for the insoles (DES= Barefoot)

Head acceleration, absolute and normalised values showed statistically significant differences for both factors and significant interaction. Barefoot resulted in the greater forehead acceleration (Figure 9.5) and participants showed quite different head accelerations, ranging between 1.24 g and 1.92 g. These results coincide with those described in Chapter 6, contradicting the hypothesis that to keep head acceleration at a safe level is the system's goal.

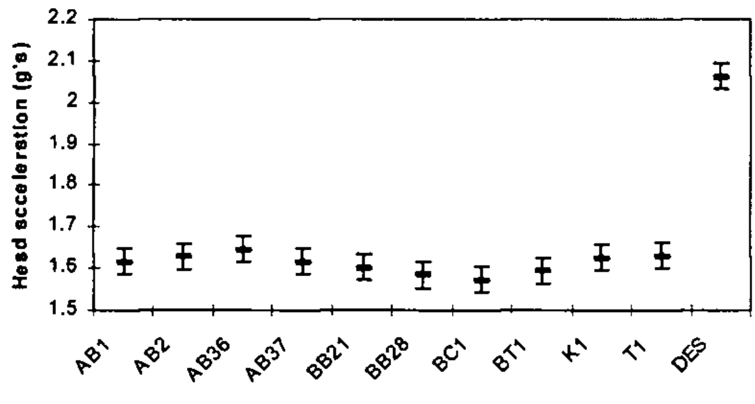


Figure 9.5. Head acceleretion for insoles (g) (DES= Barefoot).

Transmission parameters also showed significant differences. Head/tibia acceleration ratio (MC/MT) exhibited statistically significant differences between participants and between insoles (Figure 9.6). As described in previous Chapters, barefoot showed the lower transmission whilst material BB21 showed the greatest. Similar results were observed for transmission parameters when normalised with respect to barefoot.

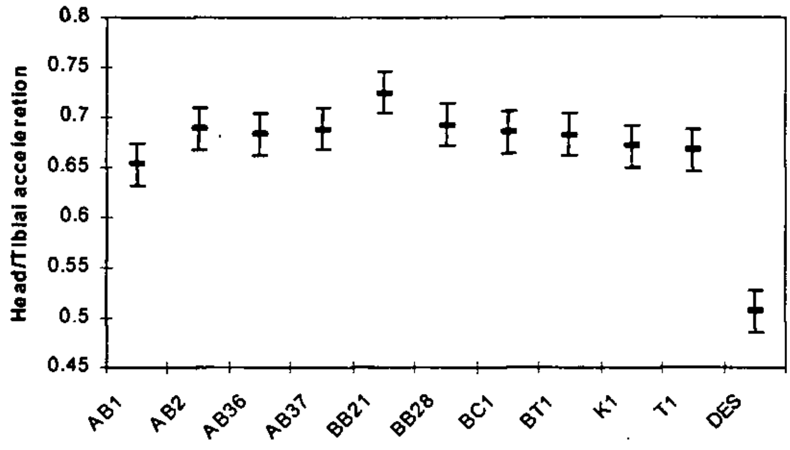


Figure 9.6. Head-tibial transmission for insoles

Transmission to tibia (MT/MF1) was different between participants and between insoles. For barefoot there were differences between participants.

The factor analysis suggests that variance of impact and transmission parameters could be explained by five principal components (Table 9.4) accounting for 92% of total variance:

i. Reduction of impact and transmission parameters with respect to barefoot

ii. Absolute parameters including negative score of head - tibia transmission (that is increase of impact parameters and acceleration was related to decrease of head transmission from tibia).

iii. Normalised tibial acceleration and transmission and negative normalised head-tibial transmission for greater acceleration in tibia and less to forehead (nearer barefoot the better)

iv. Similar to (iii) but for absolute values

v. Time to peak values (only 8%)

These results clarified data structure and were quite logical.

Insole level (CV = 0.889)

The results of Pearson's correlation analysis showed quite significant correlation (Table 9.5, only significant correlations are shown). Results at this level are obtained when the individual's variability is eliminated representing average general human behaviour when insole is changed.

Table 9.4. Factor Analysis of impact parameters at all level

	Component				
	1	2	3	4	5
MCGS_1		.584			
MCGSND_1	.845				
MCMTGS_1		-.768		-.574	
MCMTGS_2			-.942		
MF1N_1		.890			
MF1ND_1	.876				
MTGMF1_1			.923		
MTGS_1		.869			
MTGSMF_1				.969	
MTGSND_1	.546		.787		
RATAC_1	.654	.633			
RATACD_1	.812				
TMF1TI_1					.960
eigenvalue	3.368	3.201	2.813	1.537	1.040
% of Variance	25.909	24.620	21.637	11.825	8.004
Cumulative %	25.909	50.529	72.166	83.991	91.994

Table 9.5. Correlation results for insole level

	mc	mcnd	mcmt	mcmtnd	mtgs	mf1n	ratac	tmf1	mtmf1	mtmf1nd	mf1nd	ratacnd
mc	—											
mcnd	0.996	—										
mcmt			—									
mcmtnd			0.987	—								
mtgs	0.653	0.659	-0.888	-0.697	—							
mf1n			-0.668		0.655	—						
ratac			-0.829	-0.777	0.806		—					
tmf1								—				
mtmf1					0.671				—			
mtmf1nd										—		
mf1nd						0.961						
ratacnd			-0.839	-0.781	0.741		0.975					
mtnd			-0.92	-0.919	0.989	0.658	0.812		0.673			0.774

At this level significant moderate positive correlation between head acceleration and impact force and rate of loading was not observed as in the second Principal Component (in all cases lower than 0.889). Moderate correlation of head acceleration and tibia acceleration was found (0.653) as was normalised head acceleration (0.659). Tibia acceleration correlated moderately with impact force (0.655) and high with rate of loading (0.806) and normalised rate (0.741). Similar results were observed with normalised tibia acceleration.

Transmission parameters showed a moderate negative correlation of head-tibia ratio (MC/MT) with impact force (-0.668), and high and negative with rate of loading (-0.829) and normalised rate (-0.839). Similar results were observed for normalised transmission.

In this case, MT/MF1 showed a positive moderate correlation with normalised tibia acceleration (0.673), similarly though lower than at participant level. At this level, both head and tibia acceleration correlated with very high positive normalised values (>0.98) which did not occurred at other levels and Principal Components; even considering the critical value (0.444 and 0.889 respectively) this result was very remarkable. Thus an insole worn by an average participant (eliminated human's variability) showed correlation between shod and normalised accelerations.

In this case, MT/MF1 showed a positive moderate correlation with normalised tibia acceleration (0.673), similarly though lower than at participant level. At this level, both head and tibia acceleration correlated with very high positive normalised values (>0.98) which did not occurred at other levels and Principal Components; even considering the critical value (0.444 and 0.889 respectively) this result was very remarkable. Thus an insole worn by an average participant (eliminated human's variability) showed correlation between shod and normalised accelerations.

9.3.1.3. Discussion and conclusions

The impact forces and accelerations registered were in the range reported by others and those found in the literature (see Chapter 6). Impact conditions showed quite a great range of COV by just changing the insole material providing, at the same time, a sound base of stimulus for perception analysis.

Anova results reflected more differences between participants than between insoles, which was expected since individual differences due to walking pattern, free speed and other aspects are likely to be greater than differences due to insole material. On the other hand, again as expected, barefoot yielded greater impact forces and accelerations as well as greater rates of loading. However, as in Chapter 6, forehead transmission was lower for barefoot participants probably as a result of accommodation. Results with respect to barefoot were in general similar to absolute variables due to the variation of insole materials and readily explained by shod results.

Head acceleration showed significant differences between insoles [Chapter 6] Thus if keeping head acceleration at a safe value is the goal of the system, proprioceptive should be disregarded. The acceleration oscillated between 1.92 g and 1.24 g. This represents a range of about 35% so the existence of a safe threshold for head acceleration may be disregarded.

Impact forces and rate of loading showed differences with respect to barefoot but little differences between insoles when the barefoot factor was eliminated. These differences were less than 10% - around 6% for impact forces and 10% for rate of loading. Similar results were obtained in Chapter 6. This result is interesting since both parameters have been demonstrated to have a similar effect on shock absorption but, in the literature, the rate of loading has been related to impact perception. That suggests that keeping both at a safe value could reasonably be an objective of the system. Or just an indication that insole material had little influence in impact parameters.

Interesting results were obtained from principal components analysis. Five principal components accounting for 92% of total variance were identified. These components are:

i. Reduced impact and transmission parameters with respect to barefoot
ii. Absolute parameters including negative score of head - tibia transmission that supports, as stated above, that any increase of impact parameters was linked to increase of accelerations but a reduction of head transmission from tibia.
iii. A reverse normalised and (iv) absolute parameters and the relationship between tibia and forehead.
v. Temporal aspect of peak force defined an independent component that explained a low part of variance.

Impact parameters and accelerations were correlated against each other (impact force, head and tibia acceleration and rate of loading). Similar results occurred, in general, at insole level. A particularly interesting result was that when variability due to interaction between human and insole was not eliminated, the behaviour with respect to barefoot was not explained by shod results and at insole level suggesting an important human role.

9.3.2. Plantar pressure distribution

According to Robbins & Gouw (1991) plantar pressures are related to accommodation onset whilst heel pressures have been related to impact perception. In this part of the study, changes in plantar pressures were investigated with special focus at pressure shift from forefoot to rearfoot areas.

9.3.2.1. Material and methods

The aim of this part study was to register the pressure distribution under the foot during walking with different insole materials for further analysis. The IBV instrumentation described in Chapter 8 was used. The test facilities included (Figure 9.7):

- A set of instrumented insoles.
- Photocell barriers.
- Digital chronometer.
- PC for data acquisition.

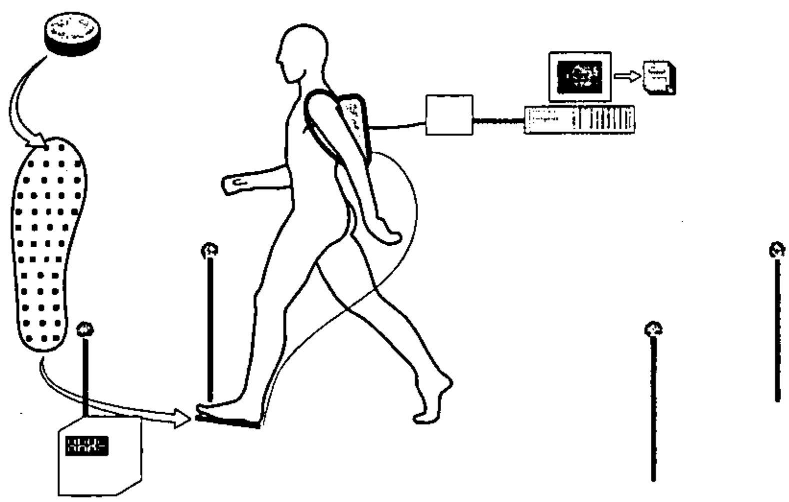


Figure 9.7. Laboratory set up for plantar pressure distribution testing.

The pressure distribution was registered with an instrumented insole placed into the shoe that, together with electronics, allowed registering pressure at each sensor at a high speed.

The digital chronometer was driven by the photocell barriers. The distance between these was fixed and the time was controlled to accept only trials at a ±10% of a fixed velocity for each participant (free speed).

Different variables were obtained from this technique, but following previous results, only the maximum mean pressure under each plant area was considered for this study. Even though the time integral of pressure has been shown to play an important role in comfort, the system used made difficult to elicit the pressure onset. Thus, this parameter was not used in this study. For the analysis, the foot plant was divided into eleven physiological zones (Figure 9.8). Pressure for each area was computed as the median of the pressure data for all the sensors in this area.

Five participants took part in this experiment. Three acceptable steps were taken for each of three acceptable trials when wearing each insole into the same type of shoe in a random order. As the tests involve insoles in shoes, barefoot testing was not appropriate.

TE	Lataral heel
TC	Central Haal
TI	Intarnal hael
PME	Lataral midfoot plant
PMC	Cantral midfoot plant
PMI	Internal midfoot plant
C5M	Fifth metatarsal haad
MM	Central metatarsal heads
C1M	First matatarsal head
RD	Toes plant
PrO	Hallux plant

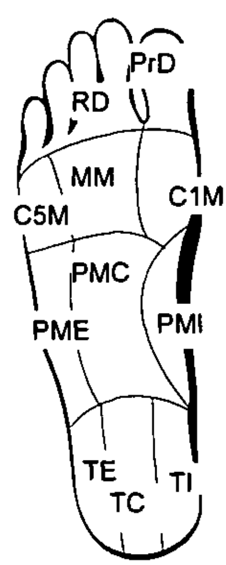


Figure 9.8. Areas into which the foot plant is divided.

The statistical analysis consisted of an Analysis of Variance (Anova) considering insole, participant and foot area as factors, as well as the second order interaction. From this analysis the influence of the insole in global pressure, of participant and insole in the pressure pattern and pressure under different foot areas were studied. A multiple range post hoc analysis was done to assess differences between homogenous groups. Assumptions for normality were tested for each variable using the Lebene test for homogeneity of variance and Kolmogorov-Smirnov (K-S) test for normality.

Regarding shock absorption, factor Analysis of Principal Components was done to ascertain the underlying structure of impact parameters and to reduce the dimension of the study for further correlation analysis with other biomechanical variables. Principal Components method using Varimax rotation with Kaiser normalisation was done at all level. Components were chosen with eigenvalue greater than 1 when rotated accounting for more than 80% of total variance. Variables score in components was considered when greater than 0.5. A minimum of five to ten cases is required for each variable included in the Factor Analysis. In this sense, at All level 100 cases were available since the aim of the study was the interaction between insole and humans, thus each combination participant-insole was considered as a case. However, this analysis was not done at insole level because of lack of power of test since at both levels only 10 cases were available. At this level, Pearson's correlation analysis was done to assess relationship between variables.

A detailed analysis of pressure patterns was not done since it was out of the scope of this thesis. Statistical analysis was done using SPSS7.5.2.s and Statgraphics Plus 2.1 for Windows.

9.3.2.2. Results

There were statistically significant differences between insole materials for the maximum global pressure (Figure 9.9). The insole BB28 yielded the lowest pressure followed by K1 whilst AB37 gave rise to the higher global pressure.

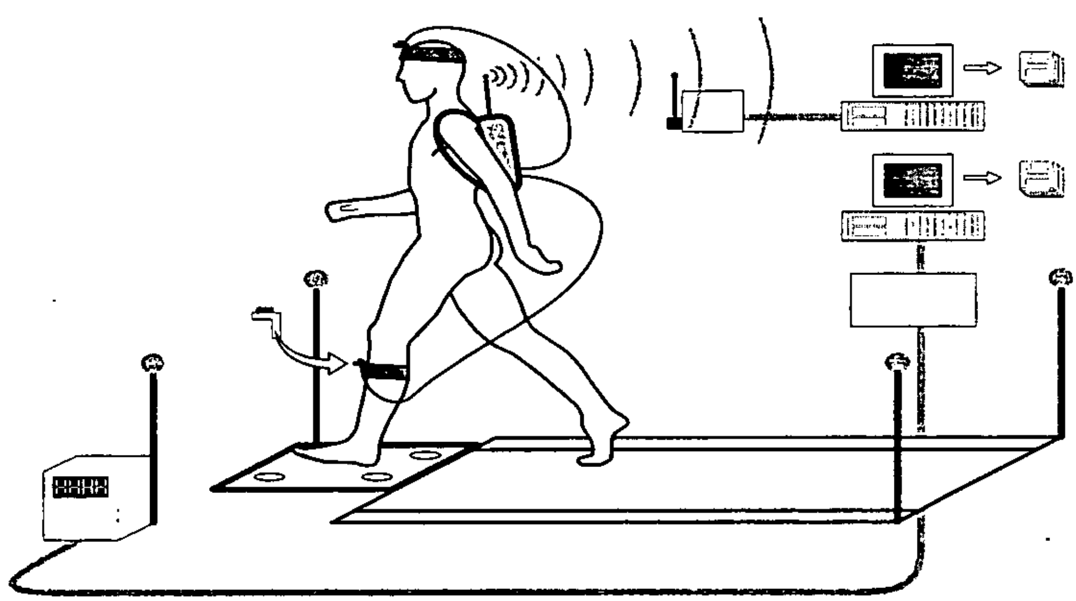


Figure 9.9. Maximum global pressure for the insoles (kPa)

The global pressure pattern for the different plant areas (Figure 9.10) was similar to previous results (Chapter 8) and those found in the literature search for males (Hennig & Lafortune, 1991; Hennig et al., 1993). The higher pressures were found under the heel, especially in the central heel (TLC), and in the central metatarsal heads (MTM), followed by the hallux (PrDD) and the first metatarsal head (C1MT). The lowest pressures were registered in the midfoot plant areas.

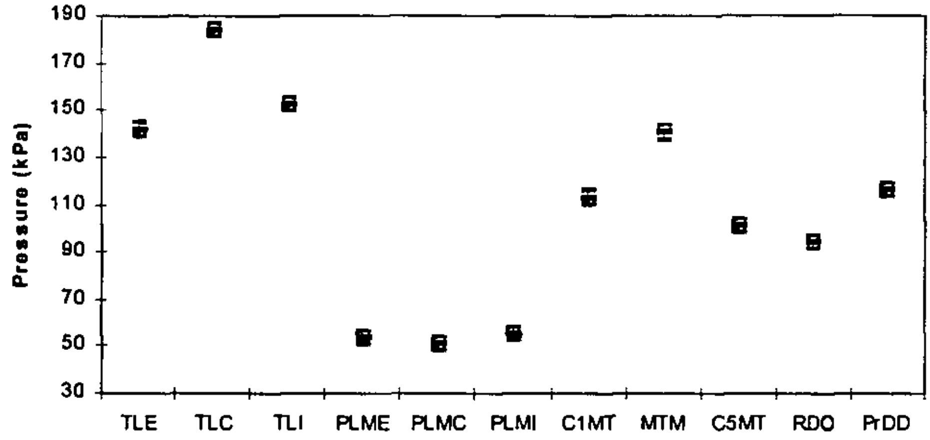


Figure 9.10. Global pressure pettern under the foot plant (kPa).

This pattern was in general common for all the insole materials, probably more related to the shoe last and construction than to insole materials. Nevertheless, some differences were observed (Figure 9.11). The figure 9.11 represents the pressure at every foot location separately for each insole. In a general sense, it would be possible to identify different patterns. Materials AB1, AB2 and AB36 showed similar distribution. The same applied for BB21, BB28 and BT1, as well as K1 and T1. In any case, a detailed analysis of pressure patterns would be quite complicated and is out of the scope of this thesis. Comparing heel and forefoot pressures according to the Robbins theory, the difference between both regions was greater for AB2 and AB37 and lower for BB21 and BB28.

The analysis of pressure under each plant area for each insole material also revealed some statistically significant differences. Figure 9.12 represents pressure for each insole separately for each foot location. The pressure under the midfoot plant areas showed little differences between materials (low pressure was registered under these areas and so the power of test was very low). However, differences were found under the other areas.

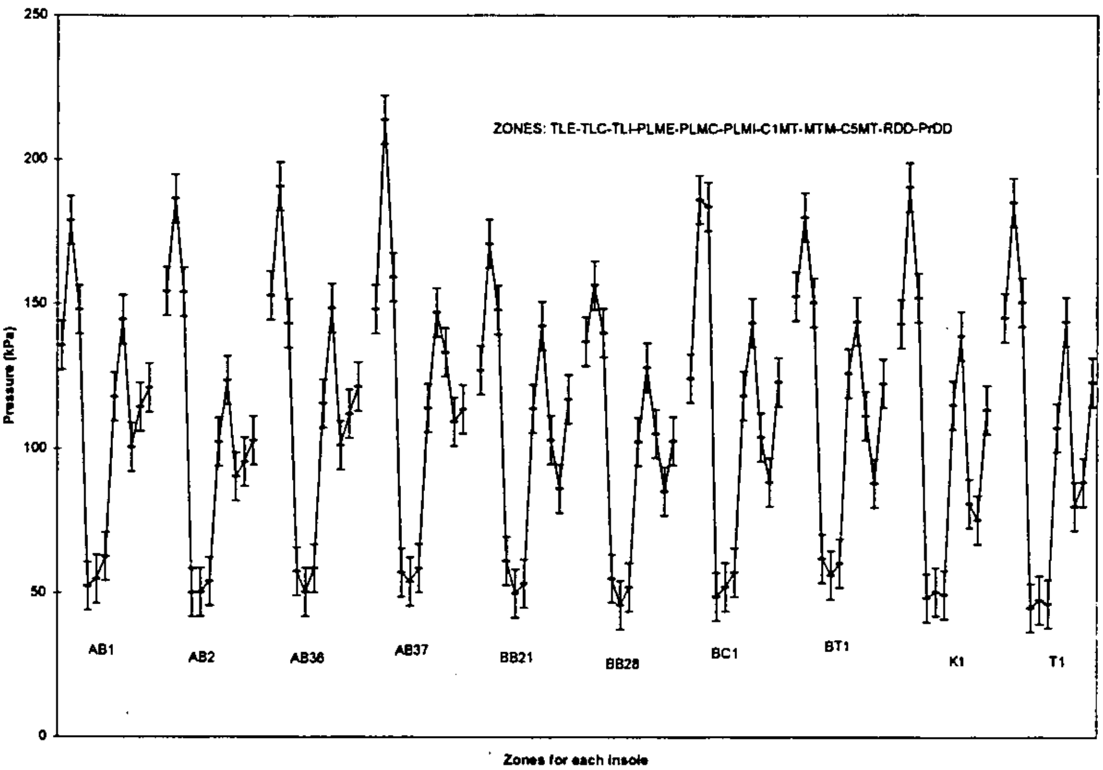


Figure 9.11. Pressure pattern for each insole material

In the rearfoot, great differences between materials were found under the central heel, and some differences under the lateral (external) heel whilst little variations were observed under the internal heel. This is important, since the literature shows (Hennig et al., 1996; Lafortune et al, 1995b; Milani et al., 1997) a relationship between lateral heel pressures and impact perception, so if different pressures were observed a variety of impact perception could be expected.

In the forefoot, great differences were found under the fifth metatarsal head and the rest of toes (RDD). Under first toe, first and central metatarsal heads little differences were observed, and mainly due to materials AB2 and BB28. At participant level, greater differences were found under all the areas except the rest of toes and midfoot plant (Figure 9.13).

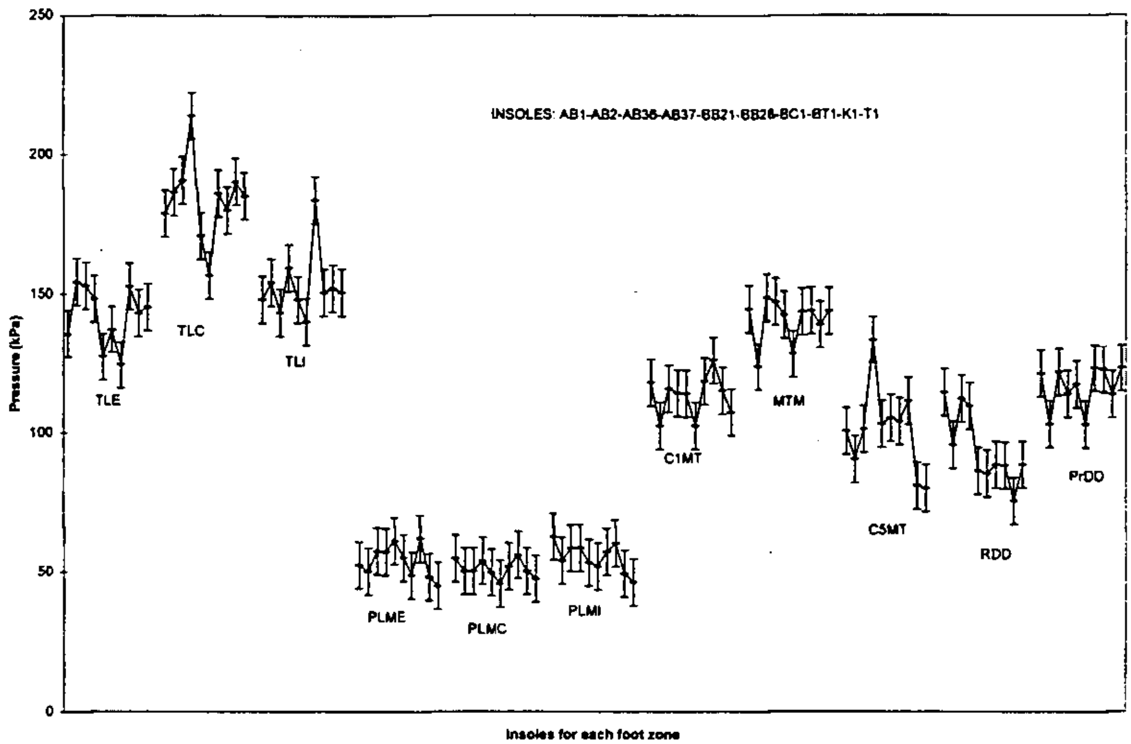


Figure 9.12. Pressure distribution at each foot plant for the different insoles

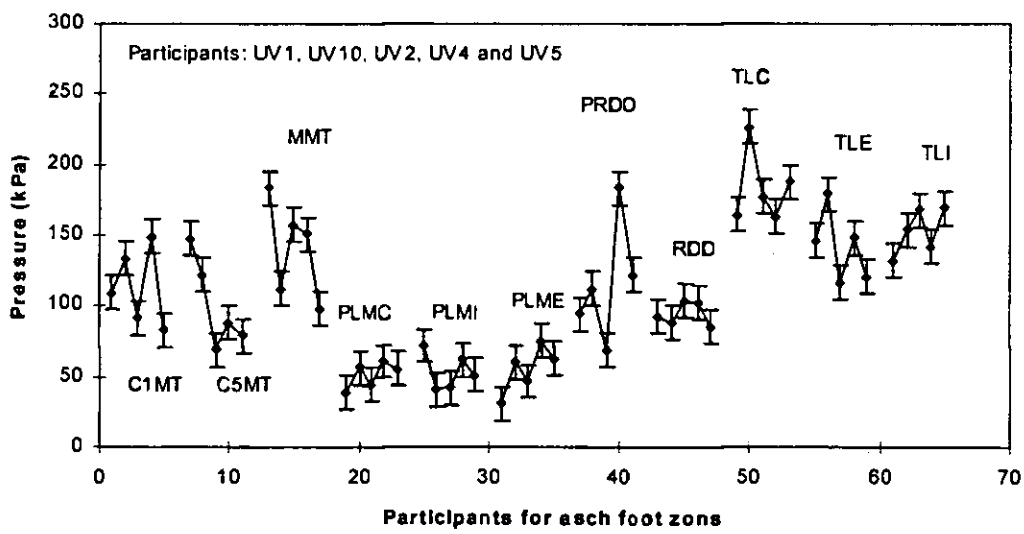


Figure 9.13. Pressure distribution at each foot plant area for the different participants

Factor analysis extracted 6 principal components (Table 9.6) accounting for more than 85 % of total variance. These components are:

i. Pressure increase under fifth and central metatarsal heads and external midfoot as well as decrease under internal midfoot plant and internal heel which could indicate a type of pressure distribution related to a more supinated or pronated foot.

ii. A component describing heel pressures

iii. Pressure decrease under central metatarsal heads and first toe increase

iv. Foot inclination into the shoe as decrease under internal heel and increase under the rest of toes were related to this component.

v. Central midfoot plant

vi. Pressure under the first metatarsal head.

Components i, ii and iii accounted for more than 50% of variance related to foot structure, heel pressure and foot inclination.

Table 9.6. Factor Analysis for plantar pressures

	Component					
	1	2	3	4	5	6
MCZ1MT						.979
MCZ5MT	.767					
MCZMTM	.533		-.591			
MCZPLMC					.859	
MCZPLME	.742					
MCZPLMI	-.792					
MCZPROD			.892			
MCZRDD				.898		
MCZTLC		.793				
MCZTLE		.879				
MCZTLI	-.530			-.515		
Eigenvalue	2.375	1.708	1.449	1.406	1.303	1.125
% variance	21.593	15.531	13.175	12.780	11.849	10.226
Cumulative %	21.593	37.124	50.299	63.079	74.929	85.155

Correlation analysis at Insole level (CV=0.889)

Insoles when worn by an average participant showed statistically significant correlations (Table 9.7), but they seemed to be related to foot structure rather than to shoe or insole effect. First metatarsal head pressures correlated high positive with central midfoot (0.808), central metatarsal (0.734) and first toe (0.755), which were both highly correlated (0.879). These areas represent the medial aspect of the foot loading structure. Fifth metatarsal head pressures correlated moderate positive with internal and external midfoot areas (both correlated), probably related to the equilibrium structure of the foot.

Table 9.7. Pearsons' correlation coefficient at insola level

	mz1mt	mz5mt	mzmtm	mzplmc	mzplme	mzplmi	mzplrdd	mzrdd	mztlc	mztle
mz1mt	—									
mz5mt		—								
mzmtm	0.734		—							
mzplmc	0.808			—						
mzplme		0.670			—					
mzplmi		0.652		0.801		—				
mzplrdd	0.755		0.879				—			
mzrdd						0.696		—		
mztlc									—	
mztle										—
mztli										

9.3.2.3. Discussion and conclusions

Three different reasons originated the analysis of plantar pressures in relation to shock absorption.

1. Impact perception has been related to external heel pressures. Differences in this value were found between materials providing a variety of stimuli for impact perception in case that literature result is confirmed.
2. Comfort is related to plantar pressure distribution that was further analysed.
3. Robbins theory considers plantar pressure shift as accommodation onset. No shift between rearfoot and forefoot pressures was observed. Insole material was showed to modify global plantar pressure. Although a general pressure pattern coincident with the literature (Hennig et al., 1991, 1993) and previous results (Chapter 8) was observed for all the insole materials, this was probably more related to shoe last and construction than to materials although some differences were observed making it possible to identify different patterns.

In the rearfoot, great differences between materials were found under the central heel, and some differences under the lateral (external) heel whilst little modifications were observed under the internal heel. In the forefoot, great differences were found under the fifth metatarsal head and the rest of toes (RDD). Under first toe, first and central metatarsal heads the differences were small. In this sense, differences under the forefoot could, according to Robbins' theory, initiate accommodation if human's tolerance is surpassed and some relationship between plantar pressure and shock absorption could be investigated.

The Principal Components identified the following six components explaining more than 85 % of total variance:

i. Mechanism of foot equilibrium or related to foot anatomy
ii. heel pressures
iii. foot inclination.
iv. decrease under central metatarsal heads and first toe increase
v. central midfoot plant
vi. pressure under the first metatarsal head.

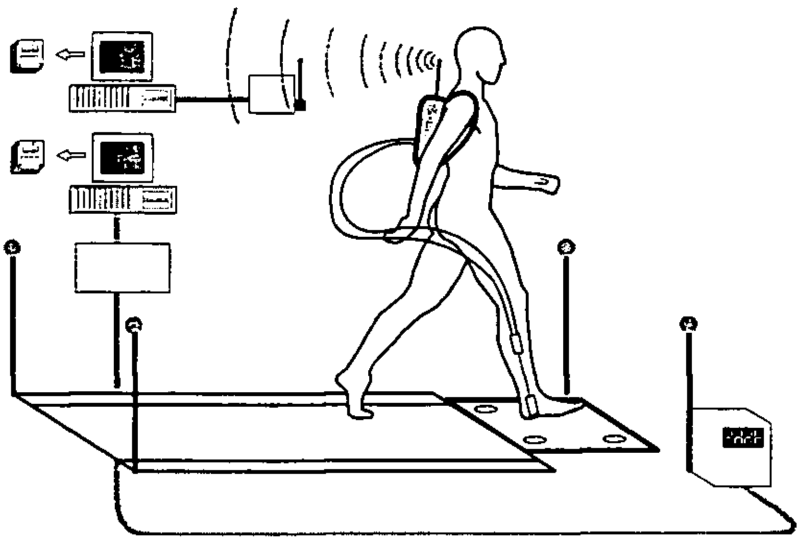
The first three components accounted for more than 50% of total variance. These components showed no accommodation but structural components were identified.
At insole level correlations were more related to foot structure: the medial aspect of foot loading structure and equilibrium function. Finally, insole material change provided sufficient differences in plantar pressures to further investigate relationship with comfort, impact perception and accommodation.

9.3.3. Rearfoot movement analysis

The aim of this part of the study was to register the kinematics of the foot and lower leg and to analyse the changes in movement in relation to shock absorption and accommodation. The literature has related knee flexion with shock absorption whilst kinematics adjustment have been ascribed to accommodation results.

9.3.3.1. Material and methods

For the analysis of rearfoot movement two different techniques were used: electrogoniometer to register ankle and knee movement and photogrametric techniques to analyse the movement of the foot with respect to the ground (Figure 9.14).



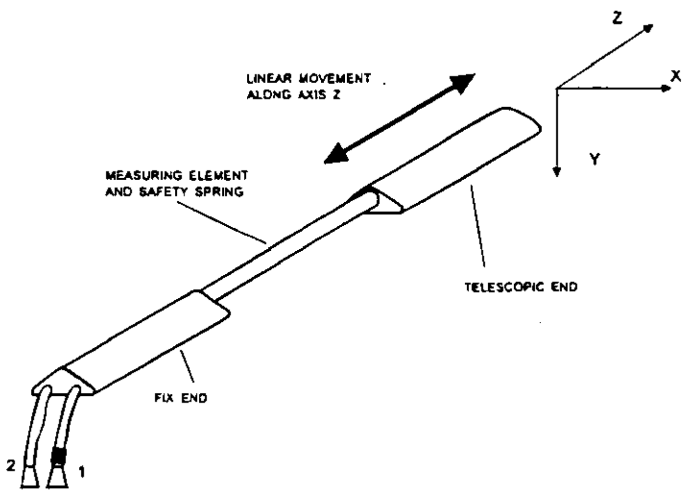


Figure 9.14. Laboretory set up for rearfoot movement analysis and Figure 9.15. Electrogoniometer

Two Peny and Gilles biaxial electrogoniometers (Figure 9.15) were attached by means of double side adhesive tape with no wrapping to avoid disturbing the foot movement. One was fixed at the lateral of the ankle with one end at the leg and the other end fixed to the shoe in such a way that the movement of the ankle was registered in two axis: flexion-extension and inversion - eversion (Figure 9.16). The other electrogoniometer was attached in the external side of the leg at knee with one end fixed to the tibia and the other fixed to the thigh in such a way that bending movement of the knee was registered.

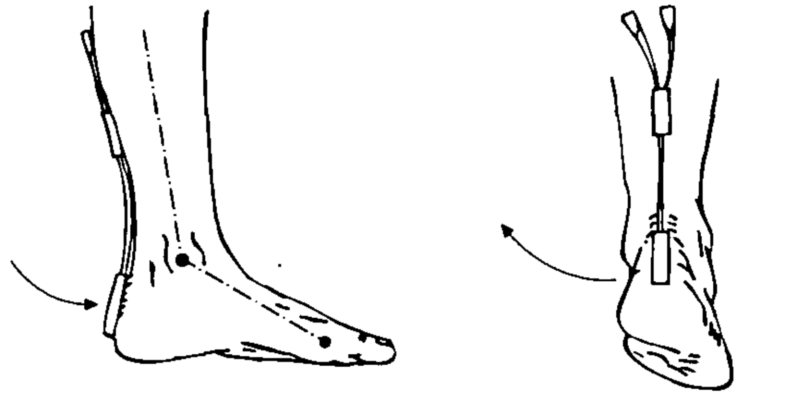


Figure 9.16. Ankle movements criteria

For the photogrammetric analysis, two reflective spherical markers were attached to the foot, one at the toe point and the other to the rear part of the heel of either foot without shoe and filmed by a video camera. A 2D photogrammetric system (IBV Kinescan®) automatically digitised the movement of the markers obtaining their x and z co-ordinates with time. These were stored for further analysis.

Five people took part in the experiment. Kinescan® photogrammes were taken of three acceptable trials under each set of test conditions for each participant in a random order. The curves of ankle, knee and foot (shoe) movements were obtained throughout the contact phase from when the foot first touches the ground until it looses contact. Many parameters were obtained from these curves, but only those related to peaks and initial contact were considered for analysis, since only the initial contact kinematics have been supposed to be related to accommodation.

The parameters considered for the analysis were:

1. At the instant of initial contact of foot with the ground:
   .- Knee flexion (FRTINI)
   .- Ankle flexion (FTTINI)
   .- Ankle inversion (ITTINI)
   .- Angle of foot (Shoe) with the ground (AATAC) (Figure 9.17).
2. Maximum value of:
   .- Knee extension (MINFR)
   .- Ankle extension (MINFT)
   .- Ankle flexion (MAXFT)
   .- Ankle inversion (MINIT)
   .- Angle of foot (Shoe) with the ground (AMAX)

As for shock absorption, factor analysis of Principal Components was done to ascertain the underlying structure of kinematic parameters and to reduce the dimension of the study for subsequent correlation and analysis with other biomechanical variables. Principal components method using Varimax rotation with Kaiser normalisation was done at all levels. Components were chosen with Eigen value greater than 1 when rotated accounting for more than 80% of total variance. Variables score in components was considered when greater than 0.5. A minimum of five to ten cases is required for each variable included in the Factor Analysis. For this, 100 cases were available at all levels. Since the aim of the study was the interaction between insole and wearer each participant-insole combination was considered as a case. However, this analysis was not done at insole level because of lack of power of test since only 10 cases were available. Pearson's correlation analysis was done at this level using SPSS7.5.2.s for Windows to study relationship between parameters and to reduce their number. Analysis of Variance (ANOVA) considering participant and insole as factors was done using Statgraphics plus 2.1 for Windows.

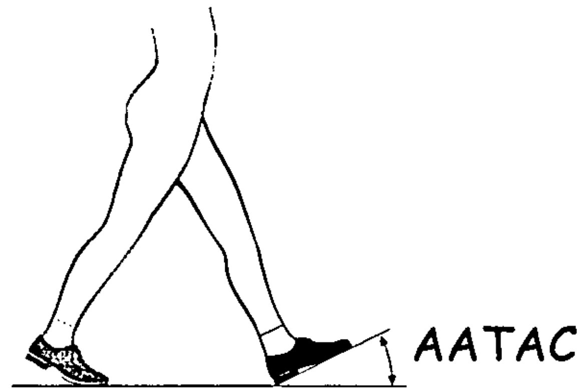


Figure 9.17. Aatac angla

9.3.3.2. Results

The factor analysis identified four principal components that accounted for more than 85 % of variance. It was interesting to note that component describing foot position and knee bending explained almost 60% of total variance (Table 9.8):

- i.- Foot position at initial contact
- ii.- Knee bending
- iii.- Ankle inversion
- iv.- Maximum ankle flexion.

Tabla 9.8. Factor Analysis for rearfoot paremeters

	Component			
	1	2	3	4
AATAC	.794			
AMAX	.840			
FRTINI		.635		
FTTINI	.938			
ITTINI			.945	
MAXFT				.916
MINFR		.839		
MINFT	.844			
MINIT			.639	
Eigenvalue	3.348	1.874	1.422	1.087
% variance	37.196	20.823	15.802	12.073
Cumulative %	37.196	58.019	73.821	85.894

The correlation analysis reflected interesting results at Insole level (CV=0.889). Similar results for the angles defining leg-foot position at initial contact were found. Parameters from each joint were correlated each other e.g. initial ankle flexion and minimum, but low, correlation with maximum ankle flexion.

Other low correlations were found between initial knee flexion and minimum ankle inversion, between minimum ankle flexion and maximum knee extension and minimum and initial ankle inversion. Also minimum and initial ankle inversion. The ANOVA was not done for Amax since it was highly correlated with Aatac and only the latter variable was used for the analysis.

Results reflected significant differences between participants as well as between insoles for many parameters. **Aatac** showed differences between participants (Figure 9.18) ranging between 17 and 35 degrees, but not between insoles.

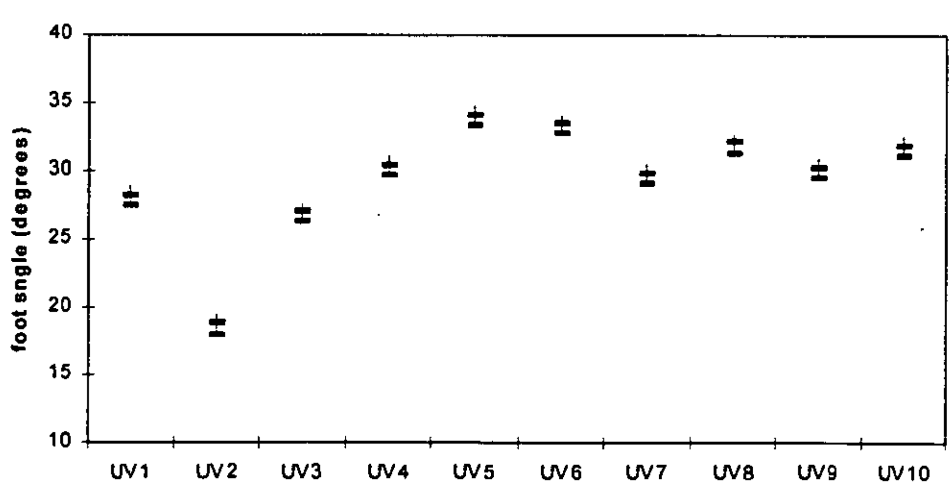


Figure 9.18. Foot engla at initial contact by participants.

Minimum knee flexion showed significant differences between participants (Figure 9.19) and between insoles (Figure 9.20); the insole differences being greater than participant's differences.

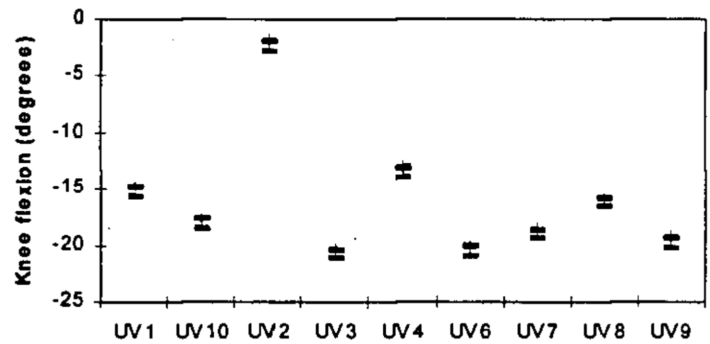


Figure 9.19. Minimum knee flexion for participants

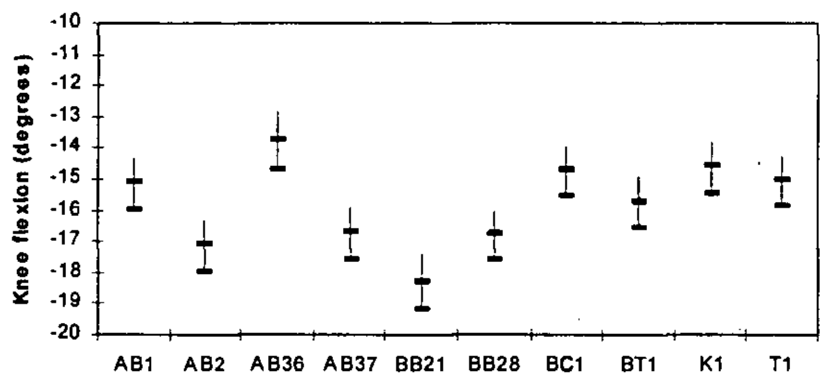


Figure 9.20. Minimum knee flexion for insoles

Similar results were observed for initial knee flexion (Figure 9.21 and 9.22), initial ankle flexion and inversion (Figure 9.23 and 9.24).

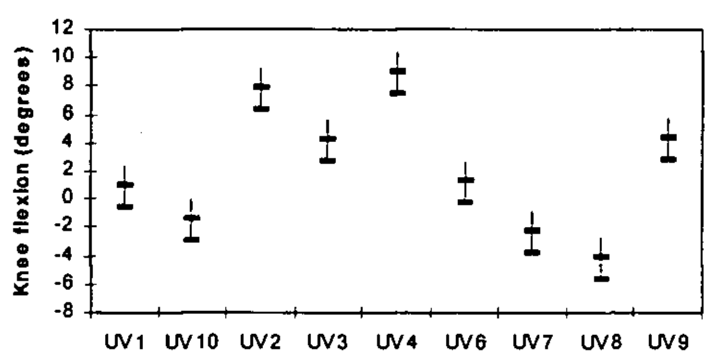


Figure 9.21. Initial knee flexion for participents

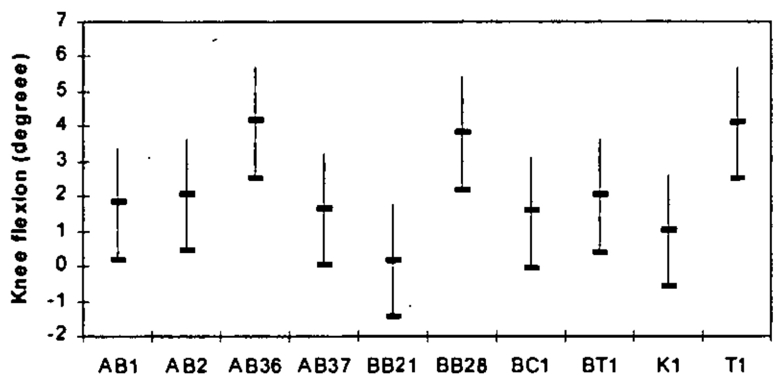


Figure 9.22. Initial knea flaxion for insolas

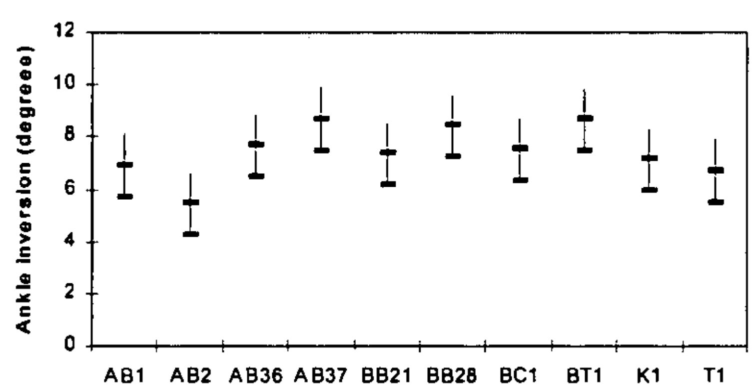


Figure 9.23. Initial ankle inversion for insoles

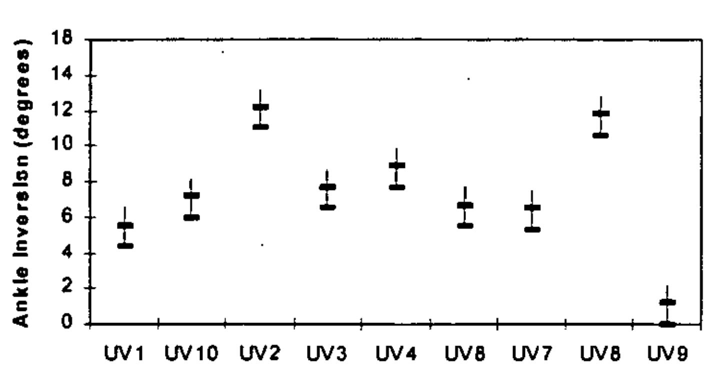


Figura 9.24. Initial ankla inversion for participants

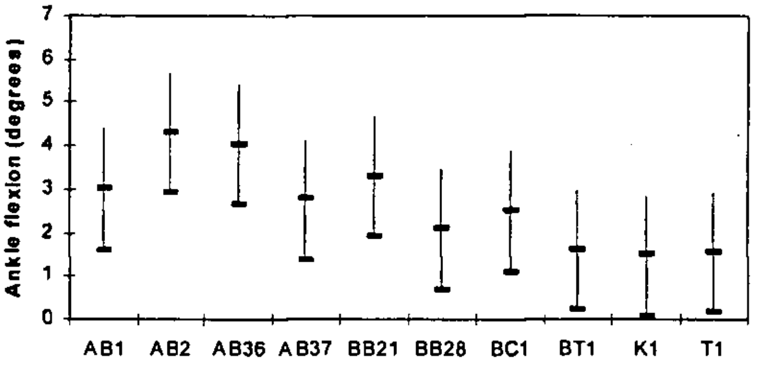


Figura 9.25. Initial ankle flexion for insoles

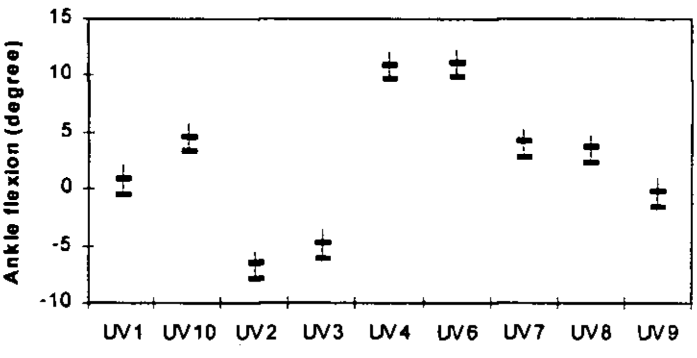


Figure 9.26. Initial ankla flaxion for participants

9.3.3.3. Discussion and conclusions

The factor analysis extracted four principal components and more than 85% of variance was explained. First of these described the foot position at initial contact, the second was related to knee bending and the third to ankle inversion movement. The forth was related to maximum ankle flexion. As a result, variance in walking kinematics at initial contact may be explained by foot position, knee movement and ankle inversion and flexion that represent patterns of movement. Foot position and knee bending explained almost 60% of total variance.

These results described a general correlation of variables defining foot-leg position at initial contact with the ground. Some correlation between minimum or maximum and angles at initial contact for ankle and knee were also observed in flexion. That was the case for insole level. Anova results reflected significant differences between insoles in rearfoot movement, which should be due to interaction. They were lower than differences between participants and seem to be related to walking kinematics.

9.4. SUBJECTIVE TESTING

Subjective testing is aimed at collecting participants' opinion and perception on a given aspect. Subjective techniques sought information about the participants' perception of impact events and system's goal.

There were two main subjective studies:

i. Comfort analysis to measure comfort level and participants' opinions when walking wearing different insoles.
ii. Impact perception under different test conditions with participants wearing a series of insoles.

9.4.1. Comfort analysis

According to the Robbins theory, the goal of accommodation is avoiding plantar discomfort. This study collected human perception of comfort and discomfort to assess the role of insole material in walking.

9.4.1.1. Material and methods

The comfort test was done following IBV methodology. There were ten male volunteers and five insoles randomly assigned to each in such a way that a minimum of five participants tested each insole. The same model of shoes was used for all insoles.

The methodology for comfort analysis has been widely used in ergonomics (Corlett, 1989; Shackel & Bishop, 1969). It was modified and adapted by the IBV for furniture analysis and since then has become a powerful and versatile tool for the development and evaluation of human related products, especially footwear. Participants are subjected to a series of subjective tests under controlled conditions and comfort perception evaluated by interviewing participants before and after the activity. These interviews collect opinions about general magnitude of comfort and discomfort in different body areas and the participants' perception of different aspects of footwear. The statistical analysis enabled the desired information to be extracted from the answers. Adapting the general procedure of Shackel (Shackel et al., 1969) and, Corlett (Corlett, 1989) two type of tests were conducted.

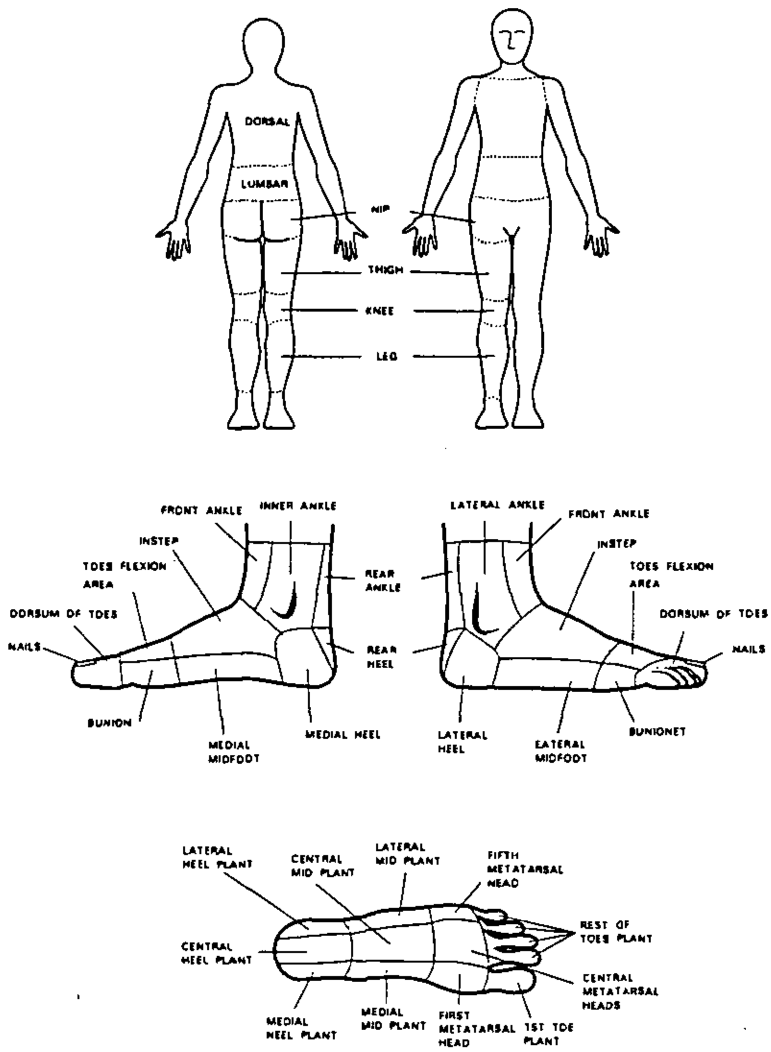


Figure 9.27: Body areas diagram.

i. General comfort. By means of a questionnaire general comfort is quantified in a 7 point Likert type scale. In this scale 1 is extremely comfortable and 7 extremely uncomfortable, whereas four is neither comfortable, nor uncomfortable.

ii. Discomfort in body areas. This test is done against a diagram of the human body divided in phyisological areas (Figure 9.27) and the participant is asked to relate the magnitude of discomfort, if any, in the different areas following a pain increasing scale of 5 points from no pain to severe pain. From this analysis, the results of discomfort in body areas it is possible to relate pain to the use of footwear. For the thesis aims, a special focus was placed on plant areas.

Subjective testing was carried out before and after one hour walking with the insoles under controlled conditions. Participants completed a questionnaire [Appendix A6] before and after using the footwear.

The initial questionnaire (before use) collects general impressions on footwear function and comfort as well as health problems. Anyone reporting significant health problem at the time of testing was excluded from that session. The final questionnaire (after use) collects comfort level and health problems caused by the shoe, including the insole effect.

Results from initial questionnaire may be considered as performance at purchase and the final questionnaire may be considered as performance in use. Comparison of these results is considered important in the analysis of the evolution of footwear properties with use.

Statistical techniques used were a descriptive analysis of data including frequency analysis, namely non-parametric Kruskal-Wallis rank analysis of variance, using insole and participant as factors and cross-tabulation to study any relationships between subjective perceptions. Statistical analysis of the questionnaires provided the following information:

i. General Comfort Index. This is a measure of comfort performance of the insoles in shoe both at purchase and in time [use]. A descriptive analysis and a rank non-parametric analysis of variance of KrusKal Wallis were done. The comparative analysis of the results for different products allowed the comfort level for a given product to be determined.

ii. Discomfort in body areas. This indicated the effect of poor footwear design on the health of users in terms of discomfort and pain in different body areas. A descriptive analysis of frequency of pain at each area was done with special focus in foot plant according to Robbins.

Finally, Spearman non-parametric correlation analysis was done to assess relationship between the perception of different aspects of footwear. As for objective tests, the analysis of results was done at three different levels: all (all human-insole interaction), insole and participant level. For insole and participant levels of analysis, the rank obtained from Kruskal-Wallis Anova analysis was used as a measure of average value for each variable as an alternative to use the mean of the scores which ranged between 1 and 5 and were not linear. Assumptions for normality were tested for each variable using Lebene test for homogeneity of variance and K-S test for normality

9.4.1.2. Results

Results are presented in order of (i) before walking (ii) after 1 hour walking and (iii) Spearman correlation analysis between subjective opinions.

The general comfort level before walking for all the insole materials presented a fairly normal distribution ranging between 1 and 4, centred in level 3 (comfortable) with a tendency to be very comfortable (mean 2.6) (Figure 9.28). The coefficient of variation (standard deviation/mean) was 24.6% demonstrating little dispersion.

The Kruskal-Wallis analysis reflected no significant differences ($p = 0.899$) in comfort level between insoles. No pain and discomfort were detected before walking otherwise the participant would have been excluded from any testing.

Initial comfort showed no significant differences between participants, although greater differences than between insoles were observed ($p = 0.161$). Participants UV2 and UV3 were the most comfortable with the insoles (conformist) whilst UV5 was the most critical

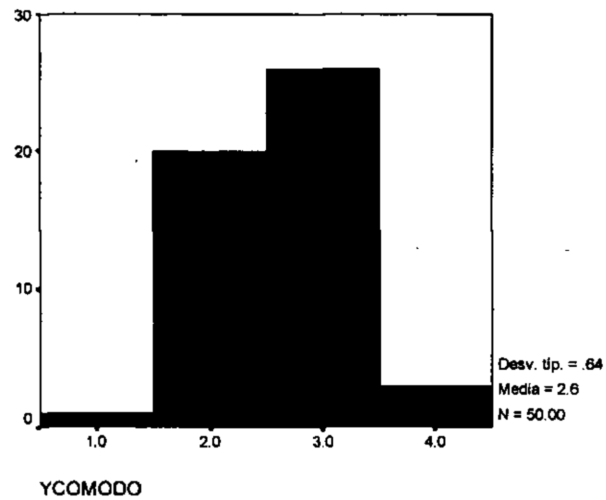


Figure 9.28. Comfort level at purchase instant

The magnitude of general comfort after one hour of use showed a slight decrease with respect to initial value, even though the distribution remained centred in comfortable (3) and the mean value remained the same (2.6), the distribution spread was wider moving towards the uncomfortable and ranging between 1 and 5. The standard deviation increased from 0.64 to 0.75 (Figure 9.29) and the coefficient of variation increased to 28.8%.

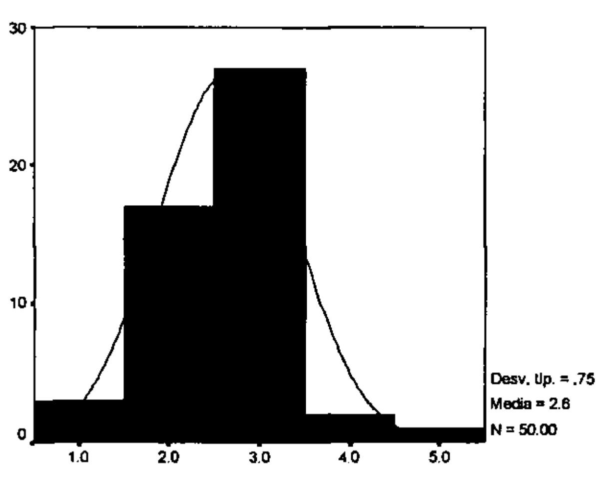


Figure 9.29. Comfort level after use

No significant differences were found between insoles ($p = 0.609$), so insole material introduced no differences in comfort level. However at participant level, significant differences ($p = 0.041$) were found after one hour walking (Table 9.9). UV3 was again the most conformist whereas UV2 changed drastically and was the most critical together with UV4 whilst UV5 was in the middle. Thus comfort seemed to depend on the individual characteristics

Table 9.9. Final comfort ranks for participants

Participants	Mean rank
UV3	12.40
UV10	16.40
UV9	20.80
UV8	23.00
UV1	23.20
UV5	25.20
UV6	29.60
UV7	34.00
UV2	35.70
UV4	36.90

Discomfort was reported at different areas of the body but with low frequency. It was interesting that the most frequently occurring discomfort was under the central metatarsal heads (8%) (Table 9.10) which could be related to Robbins theory. In general, discomfort occurred under the metatarsal heads, external midfoot plant, plant of toes and central heel plant.

Table 9.10. discomfort frequency

Body Area	Frequency (%)
First metatarsian head	4
Fifth metatarsian head	4
Central metatarsian heads	8
Fifth metatarsian lateral	4
Posterior leg	6
Lateral medial plant	4
Plant of toes	6
Central heel plant	4
Internal heel plant	4
Lateral heel plant	4

Cross-tabulation was done to analyse the distribution of pain in participants with respect to materials but no association between insole materials and participants was observed. Discomfort should be due either to the fact of walking for one hour or to shoe design (although the shoe was of high quality, leather, plastic sole, medium last, etc.). The low incidence of discomfort (< 10%) terminates the possibility of further analysis due to insufficient data for statistical analysis.

9.4.2. Impact Perception

Impact perception has been the focus of the interest of most of current investigations related to shock absorption (see Chapter 4). The study of impact perception addresses both biomechanical and methodological questions. According to the Robbins model, impact perception is related to how the human body senses - or accommodates - the input and then passes it on to the control mechanism. The analysis of perception together with data from the other experiments was combined in chapter 11 to study input and output functions of the proprioceptive system. In addition, the following methodological questions were investigated:

i. Are human beings able to perceive small changes in insole material during walking?.

ii. To compare the impact perception in active and in passive conditions. There is no general agreement [Chapter 4] on the results from active and passive testing. Literature on impact perception refers mainly to passive testing which implies that control of initial impact conditions is the main advantage and considers the absence of movement and past experience as shortcomings. Surprisingly, to date, there has been no comparison between impact and passive procedures.

iii. To check discrepancies in methodology reported in the literature with respect to reference condition, information to participants about the range of impact conditions and the way of assessing perception.

To address these issues, perception was collected from three different experiments: two in active mode under walking conditions and one in passive mode using the instrumented pendulum.

9.4.2.1. Material and methods

A common methodology was used in the three experiments, the only differences due to the testing and experimental aspects that were investigated. Two different active tests were carried out to study some methodological aspects.

Impact perception was assessed in the three tests using the same rating scale under similar test environments with similar procedures for informing participants about the tests. Perception was assessed using a Likert 7-point scale (Figure 9.30) ranging from too strong to too soft, central point (4) corresponded to nothing to note.

1 Too hard (Hard strike)	2 Very Hard	3 Hard	4 Nor hard/nor soft (modarate)	5 Soft	6 Very soft	7 Too soft (no feeling)

Figure 9.30. Scale for impact perception rating

Although the rating method for impact perception most commonly seen in the literature is the modified Borg 15 point scale, there are no advantages in using this technique: it was devised for perceived exertion. On the contrary, impact is a fast event that lasts only a few milliseconds and is perceived by mechanoreceptors in the foot skin and, probably, a system on the head. This can introduce a high variability and some offset in participants' response. At the same time, the estimation category method assumes linearity and depends on reference conditions, rating method and experimental protocol to avoid bias. In fact, the method is more dependent on protocol and reference use than in the scale itself. By contrast, Likert scales are widely used for comfort and perception studies, therefore a Likert type 7 points scale was used for consistency and for comparison with comfort studies. Moreover, in Spain it is difficult to justify any 15-point scale as semantically meaningful.

In all the tests, insoles were tested in random order and the first was assigned as reference condition, which was later tested in fifth and ninth order. Participants were always instructed about what an impact was and on the objective of the experiment. Testing environment was silent to allow the human to concentrate and reminders about the experiment objective were periodically given to the participant. An open question about opinion on the insole was also included.

Experimental procedures found in the literature require high level of concentration leading to a strong relationship between all variables. As only impact perception was required from all three tests a lower level of concentration was requirements of participants. Ten healthy males took part in these experiments. Written consent was obtained before testing.

Information about **active impact perception** was assimilated using two different experiments were carried out during walking.

1. Impact perception in a walking trial.

Participants performed the perception test after having completed comfort testing of 5 randomly allocated insoles for one hour, thus the participants became familiar with a random range of impact conditions. Each participant tested barefoot and with 10 insoles in a random order. Participants were instructed before testing about what an impact was and study objective whilst barefoot walking.

The first insole was assigned as reference condition and this was communicated to the participant. This insole was repeated in fifth and ninth place informing the participant about it to maintain its effect along the test, since it was assumed that participants would tend to take the prior condition as reference when rating a given insole.

Participants walked at their own pace on a hard surface (terrazzo) corridor in a silent environment before they were asked about impact perception in a 7 point Likert scale. The final question was the open question about insole and shoes perception. They were given as much walking time as desired before rating perception. The participants were continuously informed about the aim of the experiments. Perception of barefoot impact was collected last to avoid bias since its is the hardest condition.

2. Impact perception in the gait analysis laboratory.

In this test, perception was assessed during shock absorption analysis. The same general procedure described above was used. However, some differences were introduced mainly in relation to reference condition and to enable analyse some methodological points. This test was carried out after the previous one, so the participants were aware of the whole range of impact conditions and of the study objective. As in the above procedure, the first randomly allocated insole was assigned as reference and tested three times, but in this case participants were not informed about it.

Participants first walked barefoot and were informed about experiments' aims. They walked for a while with the first insole and performed three acceptable trials with each insole walking across the force plate following the shock absorption methodology (Chapters 6 and 9.1). Perception rating was collected in the last trial for each insole. The corridor was not of terrazzo but of a synthetic material. Participants were asked to step across a force plate and as this increased concentration requirements it might influence impact perception. It is important to note that perception was collected at the same time as the biomechanical variables that were needed for later analysis of any inter-relationship.

The Passive impact perception test was conducted using a similar protocol to the one used in the other two studies. In this case, passive perception was collected since the participants were not allowed to move and the impact was applied by means of an instrumented pendulum described previously. Perceptions were collected at the same time as the pendulum testing (Chapters 7 and 10).

Participants were asked to rate the impact caused by the pendulum. As before, barefoot was the last test condition and the reference test was the first random tested condition and was re-tested in fifth and ninth place.

For each test procedure, the reference condition was analysed to decide how best to treat it statistically. This analysis was done for each study separately, but using the same reference criteria. A Kruskal-Wallis ANOVA of perception was done for the reference condition and testing order (the reference was tested three times during the testing session) as factors to establish differences due to the testing order and the influence of reference in perception rating. Assumptions for normality were tested for each variable using Lebene test for homogeneity of variance and Kolmogorov-Smirnov (K-S) test for normality.

Barefoot condition was analysed separately as this was considered to be the most severe. Perception with respect to barefoot as well as other references were considered and analysed. The resulting Kruskal-Wallis mean rank for each insole or participant was stored as a representation of mean for further correlation analysis.

Finally, a comparative analysis between the results from three experiments was done using correlation analysis (Spearman analysis for all level and Pearson for participant and insole levels using the mean ranks as average values) as well as Freedman and t-test for sample comparison. This analysis compared impact forces and rate of loading obtained in the walking and the pendulum tests to assess whether impact conditions were the same in both cases. Anova and Freedman and t-test were used.

SPSS7.2.s for Windows and Statgraphics 2.1 were used for the analysis.

As stated in the literature, perception is a dynamic process, which depends on past experience as well as possible bias due to experimental protocol. In this experiment, literature recommendations were considered to reduce bias. Only impact perception was evaluated and in the pendulum test the participant had no control of ankle position to reduce concentration. Considering perception as a dynamic process, impact perception reported by a participant for a given condition might be described as the summation of actual perception of the current condition, the most recent past experience and the influence of anterior conditions. In this experiment, past experience depends on the individual, but as all of them tested at least five different conditions, past experience could, to some extent, be considered common to all participants. In any case, Anova was to assess the influence of order in impact perception to assess any evolution of perception with time. Influence of prior conditions in current experience was to some extent controlled by re-testing the reference condition every third test. In any case, perceptions against reference conditions and against barefoot conditions were also analysed to study any possible effect.

9.4.2.2. Results

Active Impact Perception

The ANOVA showed that the first time the reference was tested (first trial) it was always perceived as significantly softer than the rest of the trials and no significant differences were found between the second and the third tests on the reference conditions. As a result, the second trial of reference condition both descriptive and non-parametric rank Anova was considered adequate for further study. Many reports in the literature use the first condition as reference for perception scaling but whether it is perceived differently has been neither studied nor described. If it was softer, as in the present, work using it for scaling could introduce an error in perception analysis.

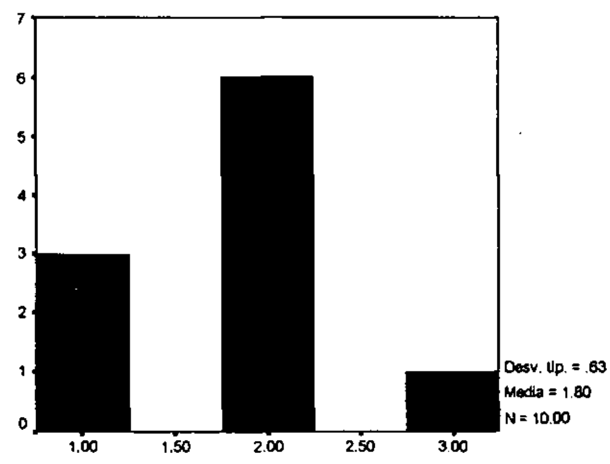


Figure 9.31. Impact perception barefoot in walking trials

The results from laboratory and walking trials reflected, as expected, that barefoot condition be always perceived as the hardest impact condition. In the walking experiment perception ranged between 1 and 3, with 2 (very hard) the most frequent score and a mean 1.8, and standard deviation = 0.63 (COV = 35%) (Figure 9.31). In laboratory

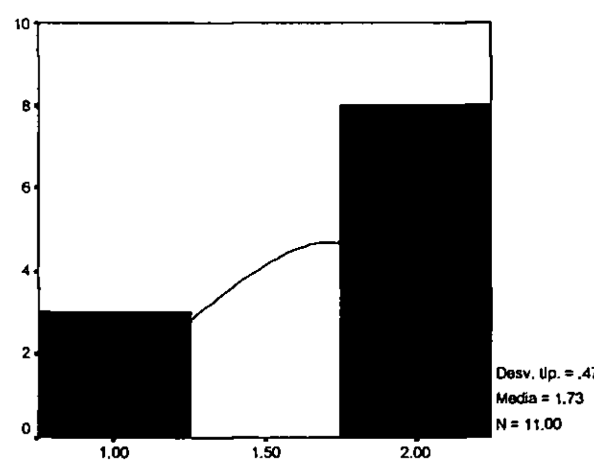


Figure 9.32. Impact percaption barefoot in the lab

test, however, harder perception and less variation were registered (Figure 9.32). The barefoot perception ranged between 1 and 2, with a mean of 1.73 and standard deviation = 0.47 (COV = 27.2%).

These differences could be due to the fact that in the second test participants were better informed since more trials had been completed with each insole and no information about reference condition was issued. On the other hand, participants needed to concentrate when stepping over the force plate.

Figures 9.31 and 9.33 showed, in both experiments, a wide distribution of impact perception between insoles. The frequency plot reflects changing perceptions due only to different insole materials in the same type of shoe. This suggests that participants effectively perceive first walking impacts and in the second walking impacts they perceived small changes in insole material.

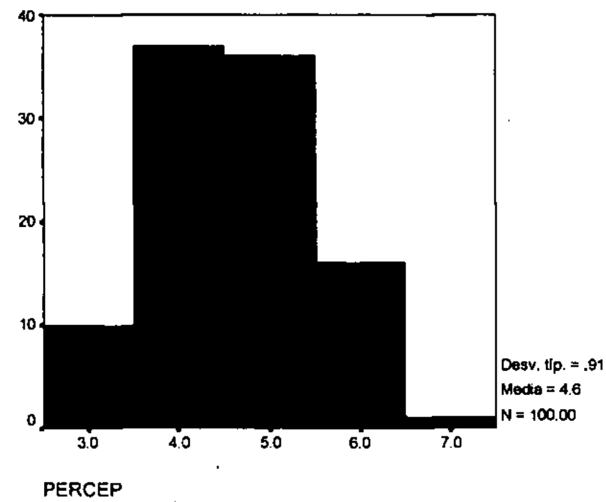


Figure 9.33. Impact perception for all the insoles in the walking trial experiment

Some differences were found between the two experiments. In the walking trial experiments, perception ranging between 3 and 7 with the mean (4.6) near to soft (standard deviation = 0.91) whilst the coefficient of variation was quite low (COV = 19.8%). It is understandable those impacts were perceived as soft since the shoe had PU soft soles. At the same time, 10% of impacts were perceived as hard (scale 3) and a much smaller percentage as soft (scale 7). Barefoot (1-3 perception) is not

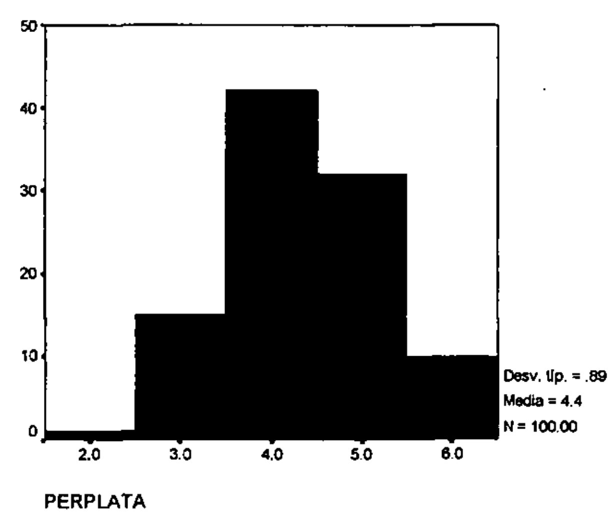


Figure 9.34. Impact perception for the insoles in the lab experiment

included in Figure 9.22

In the laboratory experiment (Figure 9.34), the mean was slightly lower at 4.4 (standard deviation = 0.89) with a similar coefficient of variation (COV = 20,2%), but impact was perceived as harder ranging between 2 and 6 which is harder (16% was hard or very hard). This compares with walking impact perception of 3 - 7). Further trials with each insole were completed to provide more information.

The Kruskal-Wallis Anova reflected statistically significant differences between insoles from both experiments although the ranking was very similar. For convenience, the results of both the walking trial and laboratory experiments are presented in Figure 9.11. From the walking trial experiment T1 was perceived as the hardest whereas K1 and BB21 were the softest ones, AB1, AB2 and AB37 tended to be hard and the rest showed intermediate values.

From the laboratory experiment (Table 9.11), T1 was also perceived as the hardest, followed by AB1. BC1 and K1 the softest. T1 and K1 coincided in the order from both experiments and the rest showed a quite coincident ranking.

The analysis of perception with respect to barefoot showed a great dispersion (COV=42%). On the other hand, assuming that the rating scale is not linear, dividing by barefoot which ranged between 1 and 3 will give rise to confounding results (it will act in turn as a magnification factor).

Table 9.11. Perception ranks for insola materials

Mean rank lab test	Material	Material	Mean rank walking trial
69.70	K1	K1	75.95
66.85	BC1	BB21	73.30
58.95	BB21	BC1	63.40
55.25	BB28	BB28	59.75
49.45	BT1	AB36	56.10
48.70	AB36	BT1	43.60
45.75	AB37	AB37	35.50
41.30	AB2	AB1	34.20
39.20	AB1	AB2	33.95
29.85	T1	T1	29.25
p=0.00	Total	Total	p= 0.00

Participant level

The analysis of perception based on participants as a factor showed no Kruskal-Wallis statistically significant differences in impact perception from the walking trial experiment, but significant differences were found in the laboratory experiment (Table 9.12). Thus, participants wearing an average insole, showed a similar perception of impacts initially, but during the laboratory testing differences emerged. So, impact perception appears to depend more on insole material than on the individual characteristics. This was contrary to comfort results, which seemed to depend more on individual characteristics. In the laboratory experiment, the participant UV6 perceived the hardest impacts and UV9, UV7 and UV8 the softest with no relationship with comfort ranking.

Table 9.12. Ranks for impact perception for participants from lab experiment.

Participant	*Mean rank lab test*
UV7	65.50
UV8	61.80
UV9	61.05
UV5	57.35
UV2	57.35
UV4	51.55
UV3	49.45
UV10	38.50
UV1	36.35
UV6	26.10
	p=.0.018

It seems that in the laboratory experiment, perception was more human dependent either because the participants had more information to judge impact perception based on past experience (the walking trial experiments had already been completed) or because they were more relaxed since they were not informed about reference condition.

The analysis of impact perception with respect to the reference condition from the walking trials showed no statistically significant differences between insoles for the quotient between reference when tested previously to every insole material and insole perception, but it did show significant differences between participants. For insoles, the mean was 1 with a standard deviation of 0.3 (COV = 30%) (Figure 9.35). However many out of range values and a high dispersion were observed (Figure 9.35). The reference condition was randomly assigned in such a way that some materials were not used as reference whilst others were referenced many times (T1 was the reference for 4 participants whereas AB1, AB2, BB21, BC1 were never used as reference). This could bias the results of this analysis. To include this, it would require a specific time series analysis that is outside the scope of this thesis, so it was not considered any further. This was also a problem in the literature survey. It is worth remarking, however, that this value ranged between 0.5 and 2.25, that is < 1 and > 1, so participants were able to judge any condition independently of the reference condition, i.e. not always either lower of greater than the reference. So it could be assumed that perception is not marked by reference.

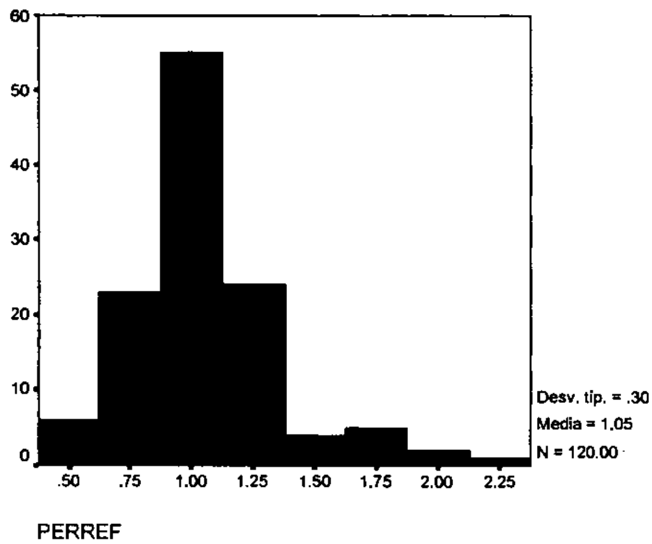


Figure 9.35. Perception with respect to the reference conditions for insole in welking trials

Passive impact Perception in pendulum tests

No statistically significant differences were found in perception of reference condition with respect to the order of test. So, to be consistent with the previous studies, the second trial for reference condition was chosen for the statistical analysis. Softer perception (mean of 4.8 and standard deviation of 0.84) was in general obtained (Figure 9.36). A narrower range was measured (between 3 and 6) with a lower variation (COV=17.5%).

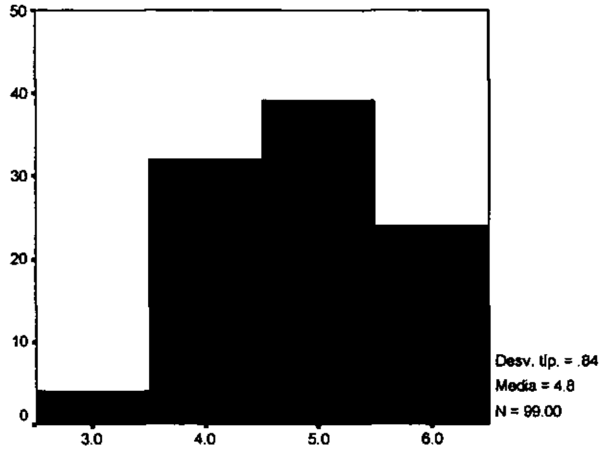


Figure 9.36. Impact perception in pendulum testing

The analysis of variance of Kruskal Wallis reflected no statistically significant differences between insoles, p=0.989. However, even though participants communicated that it was quite difficult to sense impacts when shod, statistically significant differences were found between them (Table 9.13).ç

Table 9.13. Impact perception for participants in passive study.

Participant	*Mean rank*
UV6	87.50
UV7	84.35
UV10	65.45
UV1	55.60
UV5	52.06
UV8	41.40
UV3	34.70
UV4	27.60
UV9	25.80
UV2	25.75

In this study, UV6 reported the softest impacts together with UV7, whereas UV9 and UV2 the hardest. These results contradict laboratory study results. UV6 perceived the hardest in active conditions but the softest in passive and UV9 the opposite. So, although participants were not able to perceive differences between insoles, they differ from each other in passive conditions but not in active conditions.

Comparison of perception between active and passive conditions.

The results of Spearman analysis of correlations for all data levels showed a low correlation between perception from walking trial and laboratory study (0.351) as well as with pendulum tests (0.208). A lower correlation was found between the perception from pendulum tests and laboratory experiments (0.191).

At insole level, using the Kruskal-Wallis ranks for a Pearson correlation analysis, laboratory testing showed a high correlation with walking trial (0.773) but not with pendulum test results, and no correlation was found between walking trial and pendulum. *At participant level* no correlations were found.

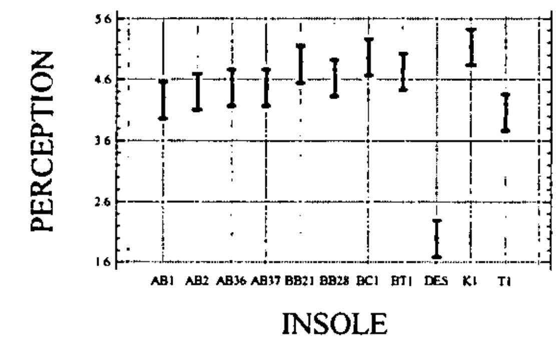


Figure 9.37. Impact perception for insoles

These results showed that impact perception with different insole materials is related in both active studies but not in the passive study. But there is no correlation between participants and study methods. This could be due to greater sensitivity of participants to testing protocol than to insoles.

Although not the indicated method for categorical data, Anova analysis was applied as it offered robustness and power to identify differences. However, it did highlight some interesting findings. Perception analysis that considered method, participant and insole as factors highlighted significant differences for all factors and significant interaction between insole and method, and between participant and method (the insole and human perception depend on the method, but not of themselves to each other). Bonferroni Multiple range test post hoc for differences between homogenous

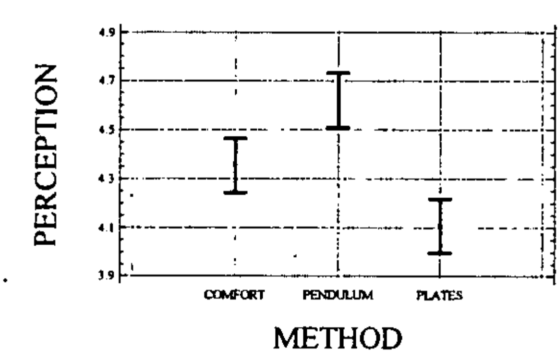


Figure 9.38. Impect perception for study methods

groups show that, as already described, barefoot was perceived as the harder and found some differences between insoles (Figure 9.37).

Differences were found also between methods (Table 9.14). Harder perception was found for laboratory results, followed by walking trials; the softest was collected from passive study (Figure 9.38). However, little differences were found between participants. UV7 perceived softer impacts than UV2, UV3, UV9, UV4, UV1 and UV8 whilst UV10 softer than UV2.

Table 9.14. Homogenous groups in impact perception for study methods

Method	LSD Mean	Group		
Lab	4.106	X		
Walking trial	4.355		X	
Passive	4.622			X

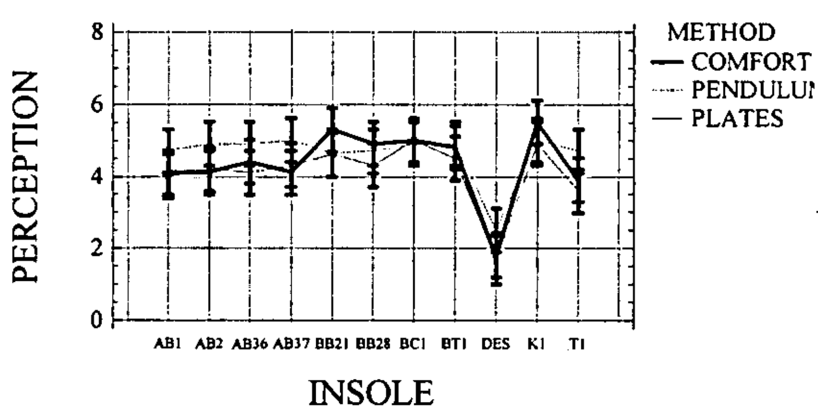


Figure 9.39. Insole-method interaction for impact perception.

The analysis of interaction found that this was significant due to some changes of magnitude for some insoles, but the general pattern did not reflected great modifications (Figure 9.39). The comparison between impact forces and rate of loading between both methods using the t-test and Freedman analysis for related samples showed significant differences for Impact force and rate of loading from active (MF1, Ratac) and passive (Fmax, Ratfi) (Table 9.15). Moderate positive correlation between MF1 and Fmax (0.665) was found. Differences in impact forces were around 15.52 N that represented less than 5%, which is considered to be very too low to be the origin of differences between active and passive impact perception.

Table 9.15. T-test for related samples for impact force and rate of loading from both methods

Couple	Mean	S. d.	standard error for the mean	COV (%)	Differences	sd	t	p
Fmax	315.24	74.06	7.16	23.49				
MF1	330.76	89.95	8.61	24.2	-15.52 (4.5%)	68.126	2.46	0.02
Ratac	17.14 10^3	14.13 10^3	1.37 10^3	82.46				
Ratfim	13.82 10^3	7.30 10^3	0.71 10^3	52.76%	3.32 10^3 (19.3%)	8.13 10^3	4.217	0.00

9.4.3. Correlation analysis between subjective and perception studies

Correlation analysis of subjective opinion of footwear properties and comfort with impact perception had the following outcomes.

At all levels, a Spearman analysis reflected no significant correlations whilst *at insole level*, using the Kruskal Wallis ranks as average value (COV = 0.894). Perception in the pendulum tests correlated negative moderately (-0.665) with initial comfort. *At participant level*, no significant correlations were observed.

9.4.4. Discussion and conclusions

No significant differences in comfort between insoles were evident but differences between individuals were found using the range of materials in this study. Similarly, little plantar discomfort was reported. Thus, there is little variation in comfort and discomfort due to changing the insole material and if accommodations did occur, it was either not due to plantar discomfort or effective enough to eliminate it.

A great variety of perception values were registered by just changing the insole demonstrating that human are able to sense impacts during walking and even small insole changes. At the same time, significant differences were found between insoles in active conditions but not in passive testing which could explain why some authors (Lafortune et al., 1995b) supposed than perception ability was limited. Differences between participants were found in both procedures.

The experimental protocol was showed to have great influence. Participants seemed to be more sensitive to the experimental protocol than insoles. Significant differences in perception scores were found between methods, these being greater between active and passive tests with no differences in impact loading in either method. This result could be due either to the fact that walking stimuli are required to perceive impacts, that is upper body vibrations are required for impact sensing or, according to Robbins et al. (1989-1991), that forefoot pressures are required for impact perception. From these results, active testing would be recommended for perception study.

Considering the reference conditions, it was observed that the condition tested initially was always perceived softer in active testing. Thus, using it as reference for scaling the rest of perceptions (as in the literature (Hennig et al., 1995b, 1996; Lafortune et al., 1995b)) could introduce an important error minimising perception in the subsequent insoles tested. However, this result was not confirmed in passive testing which is the method most commonly found in the literature. On the other hand, reference conditions were shown not to determine the perception on the following conditions. A more detailed study would be required to evaluate the influence of reference condition.

- Barefoot yielded the hardest perception in all three methods.
- No correlation between impact perception and comfort level of shoes was found and only pendulum perception was related to initial comfort (passive events).
- Active perception was observed to depend on insoles and humans.

Subjective and objective tests were conducted at the same time to record data for further analysis of research issues to test some of those hypotheses related to the research questions. The analysis of shock absorption, pressure distribution and rearfoot movement during walking provided some answers to the questions considered in this research. The results of comfort and impact perception tests focussed interest on the concept the proprioceptive model of human walking. However, impact testing of people in passive conditions was needed to investigate the shoe effect and accommodation.

10

Pendulum testing

10. Pendulum testing

10. Pendulum testing

10.1. Introduction

Previous results (Chapter 7) showed complex heel pad mechanics that may be described by three principal mechanical components. One of these describes the elastic behaviour under initial loading and was related to heel pad confinement and explained variance of time-to-peak force. From these results, a new concept called *passive interaction* was developed. Passive interaction refers to the modification of insole material and heel pad performance in use due to their mechanical coupling. As the heel compresses the insole, the contact area increases; this increase of contact area depends on the relative stiffness of heel pad and insole material - either the insole embraces the fat pad in an elastic manner increasing heel pad confinement or heel pad is deformed against the insole.

This was not evident in Chapter 6 probably because only heel inserts were tested which did not allow heel confinement. However, the literature contains references to the significant influence of the shoe on heel pad properties (De Clercq et al., 1990; Valiant, 1984).

On the other hand, for the third statement it was necessary to characterise the relative shock absorption properties of the heel pad and insole materials; this had to be done by passive testing to avoid any human adaptation.

For these tests, the instrumented pendulum, developed for this research, was used to assess the passive properties and impact mechanics of the heel pad shod with different insole materials. However, some improvements were made to the pendulum based on previous experience (Chapter 7). Additionally acceleration was measured in the leg and in the supporting wall to study shock transmission in passive conditions.

The goals of this study were:

- ➲ To study shod heel pad mechanics.
- ➲ To study shock transmission in passive conditions and the influence of underfoot materials.
- ➲ To analyse passive interaction between participants and insole.
- ➲ To compare the effects of material and heel pad confinement on passive properties.

10.2. Material and methods

The device and testing protocol were basically as described in Chapter 7, with some improvements to resolve methodological shortcomings that had became evident. Shock transmission was measured using two accelerometers and this also introduced some differences in the procedure. The following sections describe modifications to the testing protocol and instrumentation and also to experimental design and methodology.

10.2.1. Instrumentation

The improvements in instrumentation were aimed at:

1. *Increasing the precision and repeatibility of the drop position.*

This was achieved by means of an electromagnet at a fixed position to hold the oscillating part of the pendulum until it was manually swichted off for pendulum release. The electromagnet held the oscillating mass by means of a metallic part screwed into the rear end (Figure 10.1.b). It was placed at the desired position by means of a rod with two spherical joints to avoid any force transmission to the oscillating mass and thus minimise lateral movement of the pendulum. This rod was attatched to the fixation head so it was independent of a participant position (Figure 10.1.a).

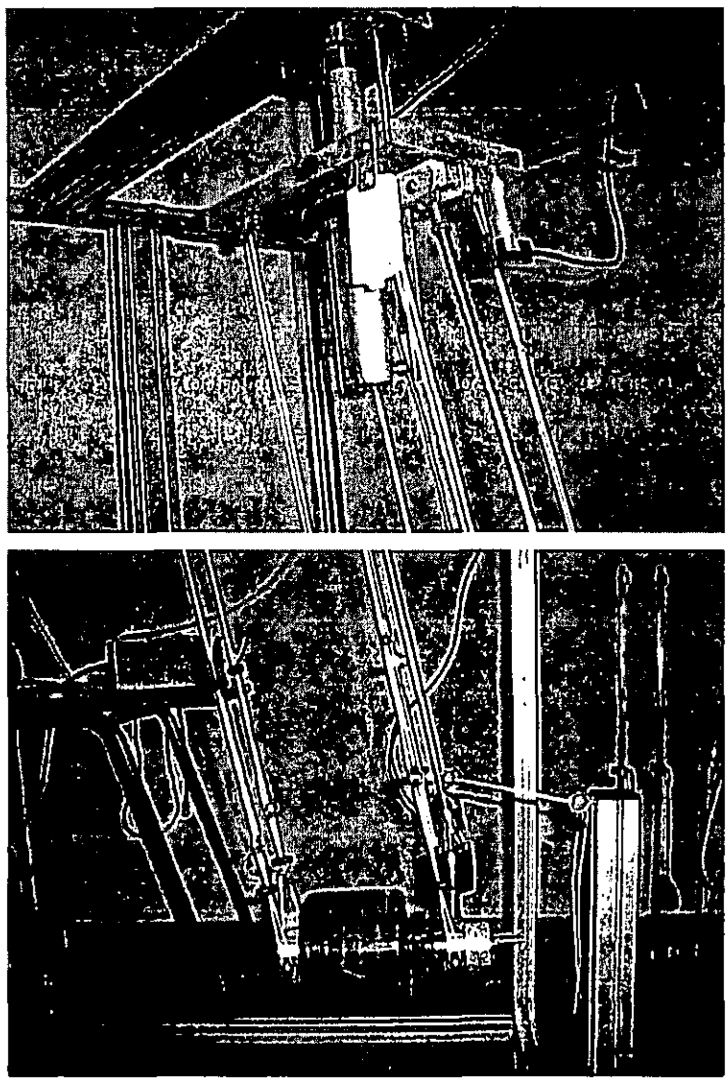


Figure 10.1. Release system a) at the fixation head, b) electromagnet holding the pendulum

2 *Increasing accuracy of heel pad displacement measurement.*

In the previous method, heel pad displacement was calculated by the double integration of pendulum acceleration and this required an estimate the impact velocity as initial condition. In this improved method, the heel pad displacement was measured using a laser transducer mounted in the leg support that could slide to adjust the position of the transducer to the participant's leg length (Figure 10.2). The transducer was placed over the participant's heel when he or she was in place. The laser beam was reflected by a plate mounted in the oscillating mass (Figures 10.3 and 10.4).

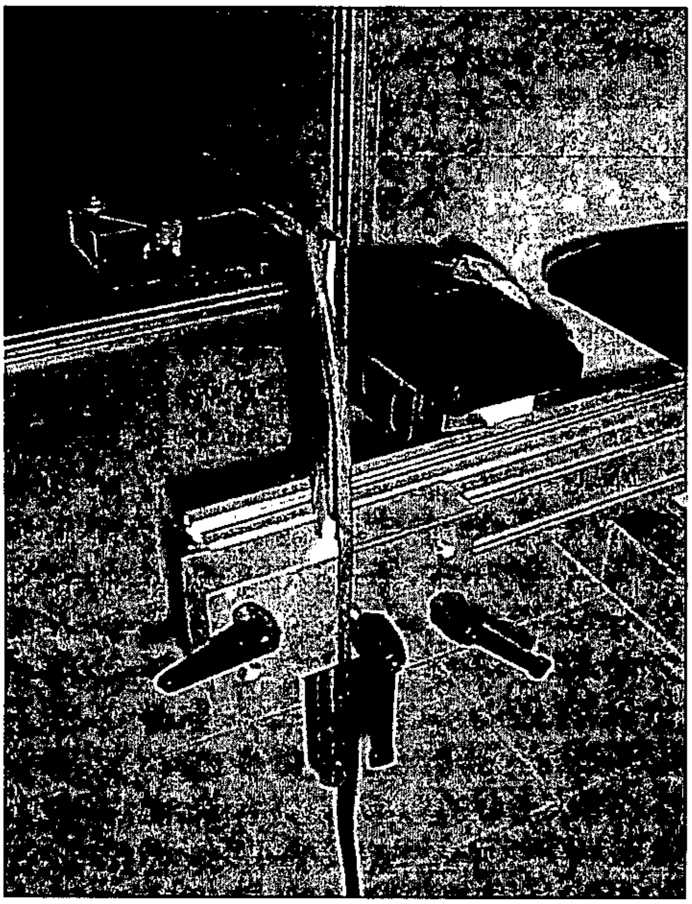


Figure 10.2. System for displecement measurement

The range and resolution of the laser were ± 2 cm and 40 μm respectively and were considered sufficient to measure heel pad displacement that is less than 15 mm (Chapter 7). More technical specifications together with the procedure for laser calibration are included in appendix A7.

3. *Measuring shock transmission.*

Two accelerometers similar to those used in laboratory shock absorption analysis were used. One was attached in the mid-third of the participants tibia by the same procedure used in walking studies (Chapters 6 and 7) and the other to the wall behind the participant support using double-sided adhesive tape. An amplifier and electronic components were needed to condition the signal from each accelerometer (Figure 10.3).

The introduction of additional apparatus necessitated more complex electronics and the connecting box was modified to receive and send signals a portable computer via a PCMCIA Data Card. This also introduced some modifications in the acquisition and processing software (explained later in this Chapter).

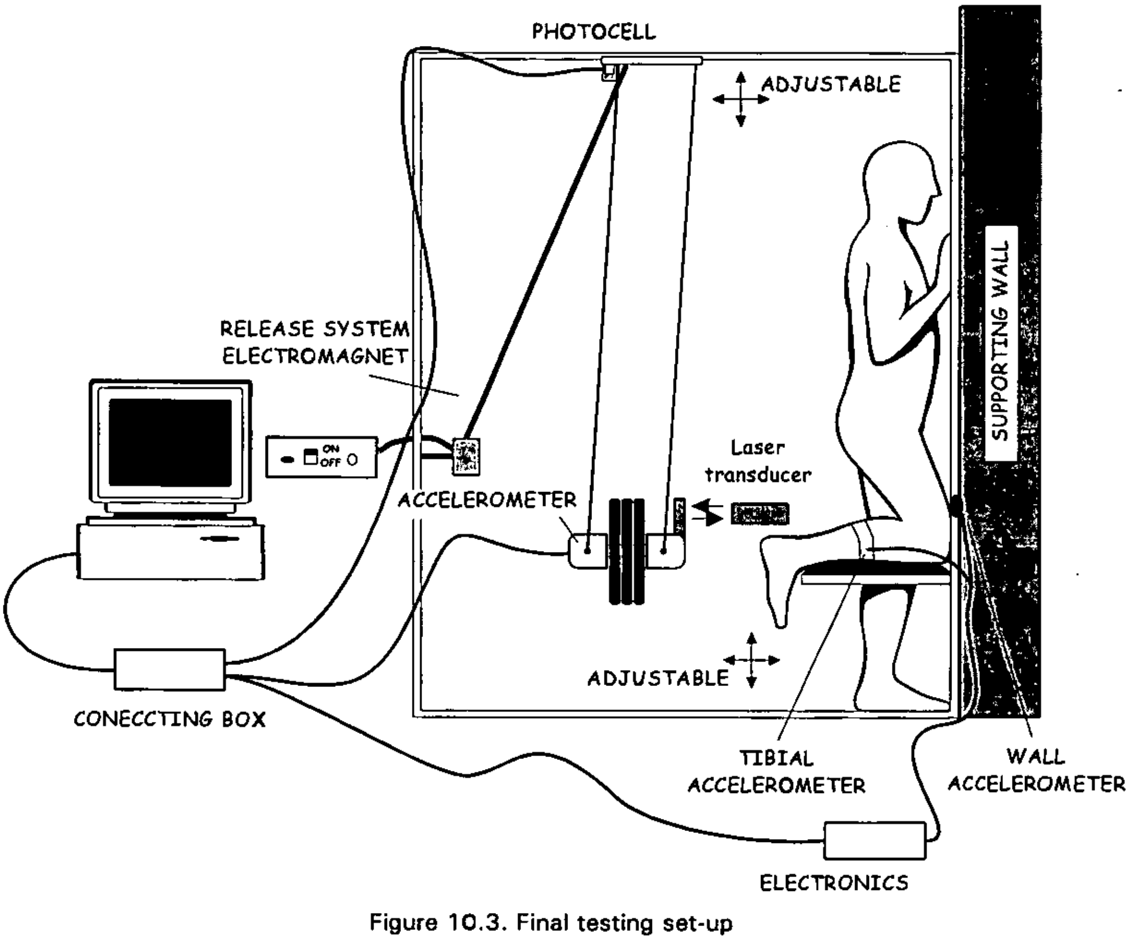


Figure 10.3. Final testing set-up

10.2.2. Testing protocol

Some modifications were introduced in the testing protocol for pendulum release, displacement measurement and to register shock transmission. Participants were first instrumented with the leg accelerometer while the test rig was adjusted for comfortable one leg standing position. Once the participant was in place, the laser transducer was positioned over the heel pointing towards the stationary pendulum. The pendulum was pulled backwards and held in place by the electromagnet. The ankle was then fixed in maximal dorsiflexion and the pendulum released without warning the participant. As in the previous method (Chapter 7) a photocell barrier/laser started simultaneous data acquisition from the three accelerometers as each was interrupted by the pendulum movement.

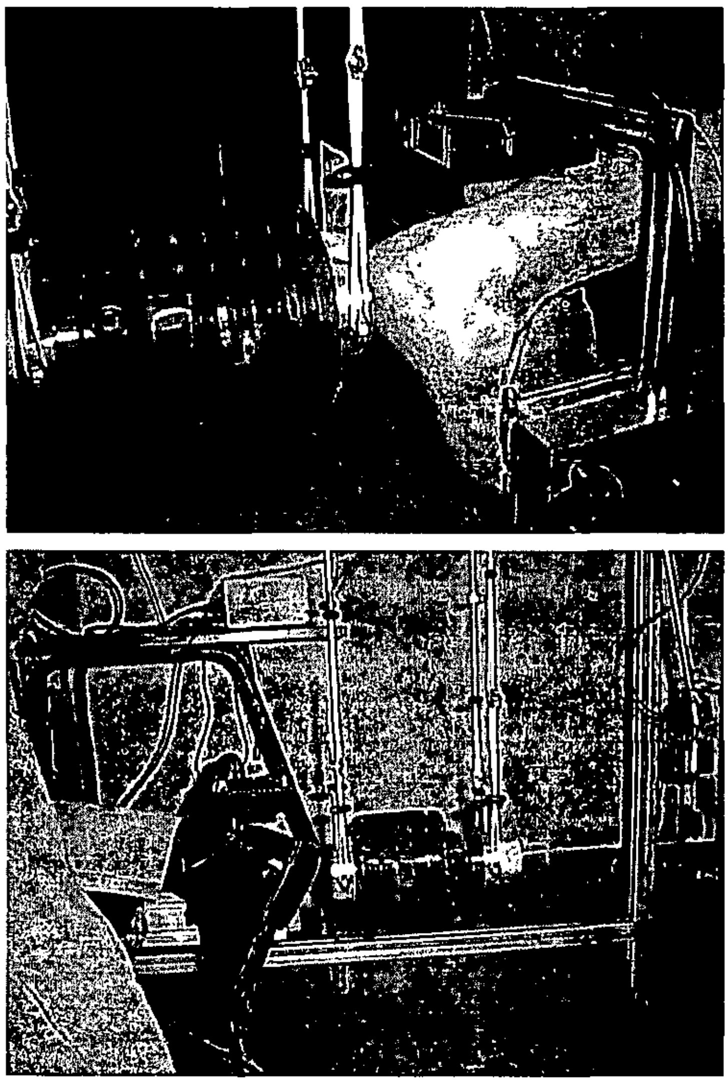


Figure 10.4. Testing shod and barefoot from different points of view.

Impact quality (impact central to the heel and without lateral movement) was visually assessed and checked in the graphic window in the computer. Ten healthy male participants took part in the experiment and each was tested in the test shoe with ten different insoles in a random order Three acceptable trials for each test condition were stored for further analysis. Barefoot test was the final test for all participants (Figure 10.4).

10.2.3. Software

Acquisition and processing software were modified as needed. All the signals were recorded electronically and simultaneously by a personal computer and a PCMCIA data card. As explained (Chapter 8) for the new machine testing method, this card reads a datum from each channel sequentially which introduced a time delay depending on the sampling frequency. That made it necessary to use interpolation techniques to translate the signals to the same time base. This was done using functions for B-Splines interpolation from Matlab library.

The modifications introduced ~~in~~ to the instrumentation and testing protocol made it necessary to change the processing software. In the new method, displacement was obtained directly from the laser transducer using the calibration curve so it was not necessary to integrate acceleration for impact velocity. Force was obtained from pendulum acceleration and mass (the mass had been changed slightly by the additional metallic part introduced for automatic release). Acceleration from tibia and wall were converted from volts to *g's* using calibration curves obtained as described [Appendix A2]. Thus the new software (Appendix A4) was simplified and parameters more precisely calculated which in turn improved the method and the quality of acquired data.

10.2.4. Data analysis

Acceleration and displacement signals were recorded at the instant of impact and stored in Matlab format (*.mat) for analysis. The signals were filtered to eliminate noise and different Matlab functions used to obtain the study parameters. These were generally similar to those described in Chapter 7, with additional new parameters to study shock transmission. Parameters can be divided into impact, force-displacement and transmission parameters.

Impact parameters *(Figure 10.5)*

.- Peak force (Fimpact), was calculated as acceleration times impacting mass, (F=ma).

.- Impulse (integration of force on time).

.- Time to peak force (TFimpact).

.- Rate of loading (Ratfim)

.- Time of impact duration (Tdura).

.- Peak displacement (Dmax).

.- Residual displacement when force reached zero value (Residu).

.- Time to peak displacement (Tdmax).

.- Time delay between peak force and peak displacement (Tvisco = TFimpact - Tdmax).

Force-Displacement parameters (Figure 10.6)

.- Energy absorption (Eabs) was calculated as the loop area by integrating force-displacement curve: Eabs = (Loading Energy - Unloading Energy) x100/ loading energy.

.- Stiffness at peak force (Kmean) was obtained as Fimpact/Displacement at peak force value.

.- Maximum Stiffness (Kmax) was the maximum on the curve of stiffness (differentiation of Force with respect to displacement).

.- Initial Stiffness (Kini) was obtained by linear regression in the region of the stiffness curve prior to maximum stiffness.

.- Final stiffness (Kfinal) was obtained by linear regression in the region of the stiffness curve between maximum stiffness and the peak force.

.- Time of shift from initial to maximal stiffness region (Tcodo).

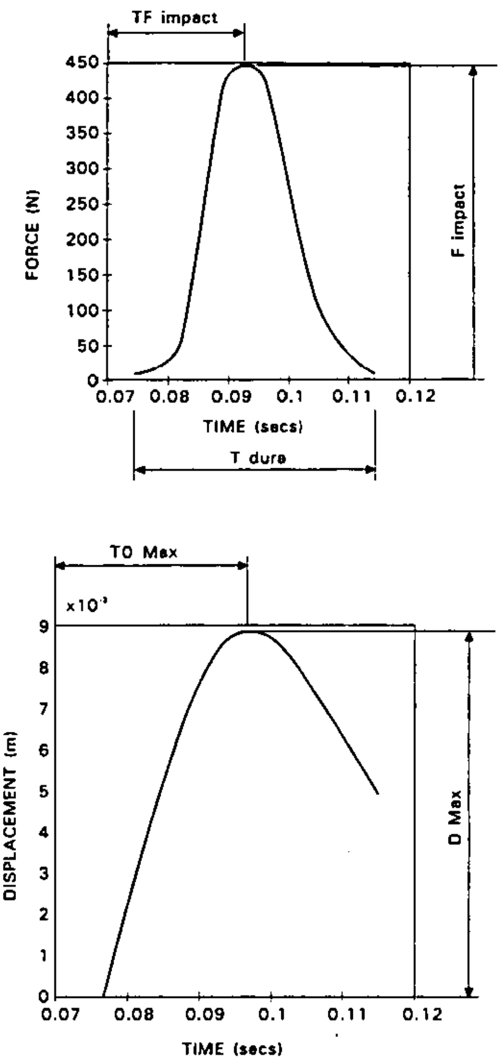


Figure 10.5. Impact parameters

Transmission parameters

.- Tibia peak acceleration (Atmax).

.- Wall peak acceleration (APmax).

.- Tibia acceleration/impact force ratio (Ratibia).

.- Wall/Impact force ratio (Rapar).

.- Wall/Tibia acceleration ratio (Ratipar).

International units were used. Force was recorded in Newtons, displacement in metres, stiffness in Newtons/metre, acceleration in g's and time in seconds.

As in Chapter 7, the following parameters were studied, as these were the basic parameters most commonly found in literature.

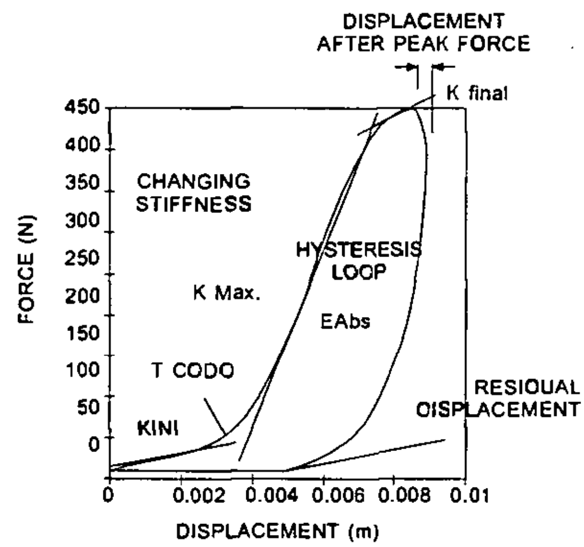


Figura 10.6. Force-displacement parameters

- peak force (FImpact)
- rate of loading
- time-to-peak force (TFimpact)
- peak displacement (Dmax)
- time-to-peak displacement (Tdmax)
- energy absorption (Eabs)
- maximal stiffness (Kmax)
- initial stiffness (Kini)

The time delay between peak force and peak displacement (Tvisco) was also studied as this is related to viscoelasticity. Time of shift from initial stiffness to maximal stiffness region (TCodo) was calculated to analyse duration of low stiffness displacement. Final stiffness (Kfinal) and residual displacement (Residual) were computed to study bottoming out. Stiffness at peak force (Kmean) was used to describe relationship between force and displacement at peak force, whilst time of duration (Tdura) was included to study the changes in the time taken for momentum transfer during impact, measured by variable impulse, which is a mechanism of impact force reduction.

Peak tibia (Atmax) and wall acceleration (APmax) as well as the ratio between them and peak forces (Ratibi and ratipa) and each other (Ratipar) were registered to analyse shock transmission. Parameters normalised with respect to barefoot (Variable/ variable barefoot) were also studied to analyse insole effect, as in the walking tests.

Because of the modifications that were introduced it was necessary to validate the new method. Force-displacement parameters were obtained using the previous (double integration) and the new procedures. Force displacement curves and energy absorption values were analysed and the results compared. The curves were similar, as was the energy absorption data. The repeatability of the new method was assessed by examining the Coefficient of variation (COV = sd/mean) of some parameters – this was low confirming high repeatability. Special attention was focussed on comparing barefoot with respect to shod.

According to results reported in Chapter 7, the mechanical behaviour of the heel pad may be described by three principal components. Consequently, Factor Analysis of Principal components was done - excluding barefoot - using Varimax rotation with Kaiser normalisation to analyse impact mechanics of shod heel pad. These components were used for further analysis. A similar analysis was done with impact and transmission parameters to clarify the impact mechanics of heel pad shod and barefoot. Force, acceleration and displacement curves were compared shod and barefoot.

A minimum of five to ten cases is required for each variable included in the Factor Analysis. At All level 100 cases were available and since the aim of the study was the interaction between insole and wearers each combination participant-insole was considered as a single case. However, this analysis was not done at insole level because of lack of power of test because at both levels only 10 cases were available. Principal components were used for the study at all levels, however due to lack of data, variables most representative for each Principal component were identified and used for correlation analysis at participant and insole levels.

Statistical analysis was done at three study levels: All, insole and participant. Descriptive statistics for all the parameters was done for each group using SPSS 7.5.2.s for Windows.

Multifactor Analysis of variance (Anova) considered insole and participant as factors, at second level interaction. Bonferroni post hoc multiple range analysis was done to study differences between homogenous groups. Assumptions for normality were tested for each variable using the Lebene test for homogeneity of variance and Kolmogorov-Smirnov (K-S) test for normality.

Pearson's Correlation analysis enabled the study of impact mechanics and shock transmission between impact, transmission and heel pad properties at the three levels

10.3. Results

The impact force, peak displacement, energy absorption, stiffness and different time parameters obtained barefoot were in the range of results previously reported (Table 10.1) and in literature (Chapter 7). Shift time and time-to-peak force were slightly longer whilst residual displacement (which had not been previously measured in the course of this research) ranged between 1.4 mm and 7.7 mm.

Teble 10.1. Barefoot results

	Mean	Maximum	Minimum	S. Deviation
APMAX (g)	.12	.16	.09	.02
ATMAX (g)	3.165	5.368	.711	1.24
Dmax (mm)	8.511	10.315	6.285	1.14
Eabs (%)	72.088	96.931	33.632	18.02
Facslas	2.73	4.01	1.30	.83
Facini	.62	3.26	-.69	1.01
Facvisco	.57	3.10	-1.94	1.50
IMPULSE (J)	8.15	8.77	7.59	.38
Kfinal (kN/m)	74.77	102.38	39.20	19.48
KFMAX (kN/m)	66.98	88.83	47.21	12.40
Kini (kN/m)	30.76	75.02	14.04	16.10
Kmax (kN/m)	137.44	207.65	82.53	32.67
RATFIMP (N/s)	$35.75\ 10^3$	$44.00\ 10^3$	$23.05\ 10^3$	$6.41\ 10^3$
RESIDUAL (mm)	4.07	7.70	1.31	1.82
Tcodo (s)	.007	.016	.003	.004
Tdmax (s)	.018	.022	.014	.002
Tfimpact (s)	.015	.019	.012	.002
Tvisco (s)	-.003	-.001	-.008	.002
Fimapct (N)	539.37	648.16	428.20	65.18

Table 10.2. Shod results

	Mean	Maximum	Minimum	S. Deviation
APMAX (g)	.046	.08	.03	.01
ATMAX (g)	.93	2.26	.26	.42
Dmax (mm)	11.14	13.44	9.25	.90
Eabs (%)	61.80	83.80	29.86	10.36
Facslas	-.28	.55	-1.64	.43
Facini	-.06	2.22	-1.49	.97
Facvisco	-.06	2.46	-2.20	.92
IMPULSE (J)	9.12	9.64	8.40	.25
Kfinal (kN/m)	26.93	41.25	6.80	5.18
KFMAX (kN/m)	27.27	34.29	18.47	3.33
Kini (kN/m)	15.65	34.81	.00	8.24
Kmax (kN/m)	40.067	89.82	26.62	9.810
RATFIMP (N/s)	$11.65\ 10^3$	$16.07\ 10^3$	$7.43\ 10^3$	$1.83\ 10^3$
RESIDUAL (mm)	3.53	6.92	1.06	1.12
Tcodo (s)	.006	.029	.000	.009
Tdmax (s)	.030	.037	.025	.002
Tfimpact (s)	.026	.033	.020	.003
Tvisco (s)	-.004	.000	-.009	.002
Fimpact (N)	293.62	340.91	235.30	21.29

The coefficient of variation for different parameters showed a good repeatability of the method (COV < 20%) at participant level but worse for insole level as expected (Table 10.3).

Table 10.3. Coefficient of veriation for study variables

	Participants	Insoles
FImpact	0.04-9.8	5.8-12.1
Eabs	0.5-23.6	12.2-25.0
ATMAX	3-9-18.9%	35.8-52.8
Dmax	2.2-9.2	5.4-13.4
Kmax	2.0-12.8	12.6-30.2

10.3.1. Principal Component Analysis

From the results of factor analysis of principal components, three principal components accounting for around 90% of total variance were extracted (Table 10.4) similarly to barefoot. The contribution of variables to each component after rotation (Table 10.4) showed some differences with respect to barefoot. An elastic component related to displacement and final stiffness, not related to maximal stiffness, was observed along with a second component that included initial elastic behaviour (Kini and Tcodo) and maximal stiffness. The residual displacement that had not been measured previously contributed to the viscoelastic component whilst delay time (Tvisco) showed a low contribution to initial - maximal component.

Table 10.4. Principal Components results for heel pad properties.

	Component		
	Elastic deformation	Initial-maximal	Viscoelastic
DMAX	-.750		
Eabs			.871
Kfinal	.870		
KFMAX	.977		
KINI		.793	
Kmax		828	
RESIDUAL			.938
TCODO		.911	
TDMAX	-.950		
TVISCO		.594	-.762
Eigenvalue	3.446	2.868	2.647
% of variance	34.46	28.68	26.47
cumulative %	34.46	63.11	89.65

In contrast to Chapter 7, the three components explained a similar part of variance. The mechanical properties of heel pad shod were different from barefoot mainly at initial loading - as observed in the force-displacement curves - in a way that principal components had a different meaning.

Principal component analysis for impact and transmission parameters (Table 10.5) indicated that two components could explain the 88.8 % of total variance. The first that accounted for nearly 60 % of variance was related to impact and wall acceleration related parameters, both absolute and normalised, whilst the second the second described a relationship between tibia acceleration and impact rate of loading, again absolute and normalised. This showed that impact forces and time were related to wall parameters whilst tibia acceleration and loading rate appeared to represent the same mechanical behaviour.

Table 10.5. Principal Components results for impact and transmission parameters

	Component	
	Impact-wall	Rate-tibia
APMAX_1	.939	
apmaxndes	.896	
ATMAX_1	.550	.757
ATMAXND		.877
Fimpact_1	.953	
FimpactND	.827	
RATFIM_1	.948	
RATFIND	.925	
RATIBI_1		.929
RATIBND		.918
TFimpact_1	-.929	
TFimpactND	-.816	
Eigenvalue	7.029	3.619
% variance	58.577	30.160
Cumulative %	58.577	88.737

10.3.2. Anova

The comparison between shod and barefoot showed, as expected, greater and higher velocity impacts barefoot, but similar patterns for force-time (Figure 10.7) and displacement time (Figures 10.8 and 10.9).

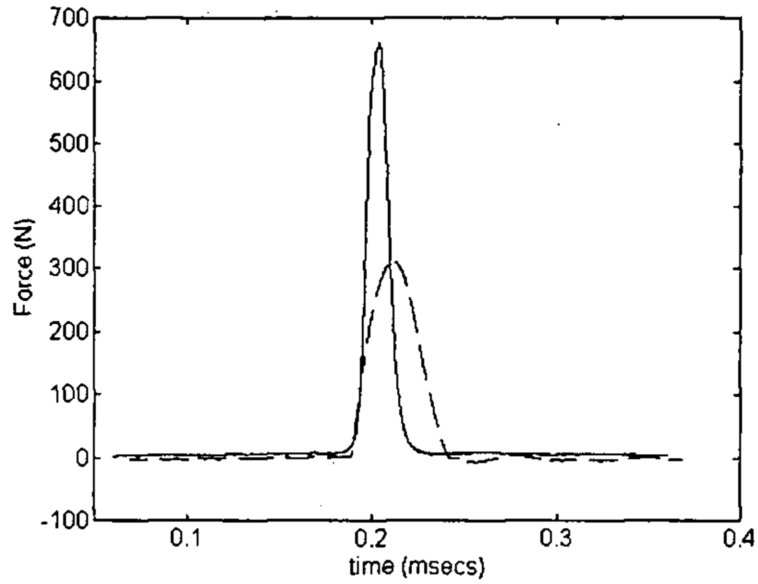


Figure 10.7. Impact curves for barafoot (continuous) and shod (dashed) for a participant.

However, acceleration signals showed a different pattern both at tibia and wall locations (Figures 10.8 and 10.9). With shod, the curves presented two minima whilst only one minimum was observed barefoot.

The force-displacement curves were different only in the initial loading and maximal stiffness areas whilst the shape of the rest of the curve was quite similar (Figures 10.10 and 10.11). The initial loading barefoot showed a longer initial phase of low stiffness with an abrupt transition whilst shod showed a smoother curve with a very short initial phase.

The ANOVA results indicated many significant differences both between insoles and between participants. Participants UV1 and UV2 were eliminated from the interaction study due to some anomalous data with some materials.

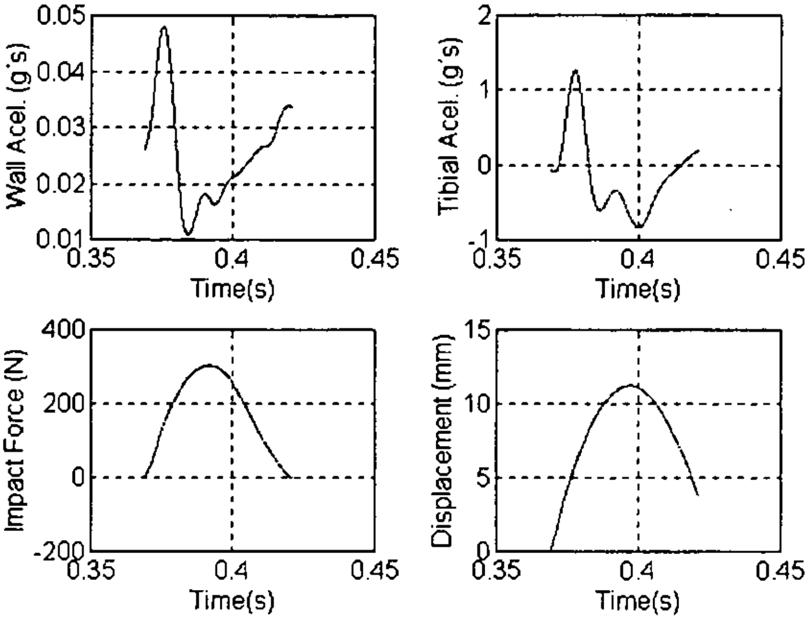


Figure 10.8. Impact force, displacement and acceleration curves shod

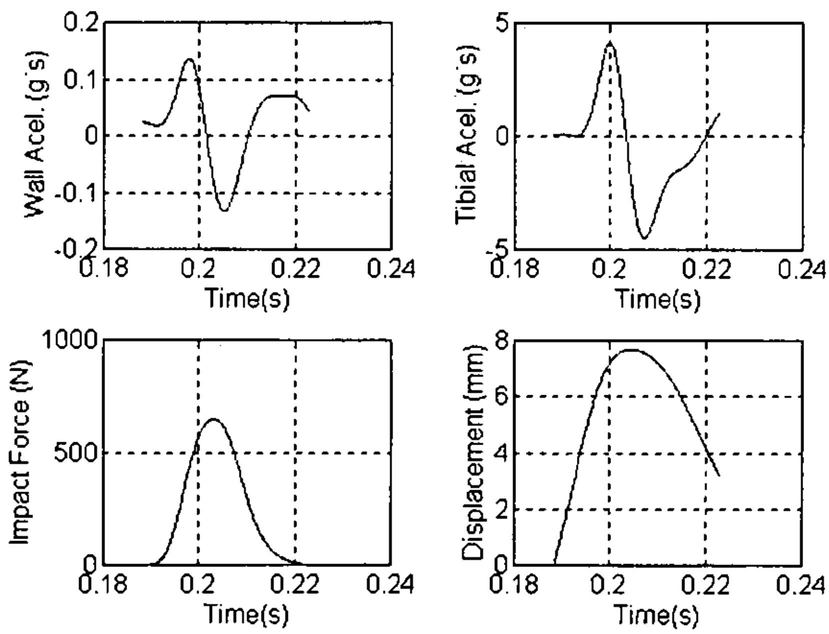


Figure 10.9. Impact force, displacement and acceleration curves barefoot

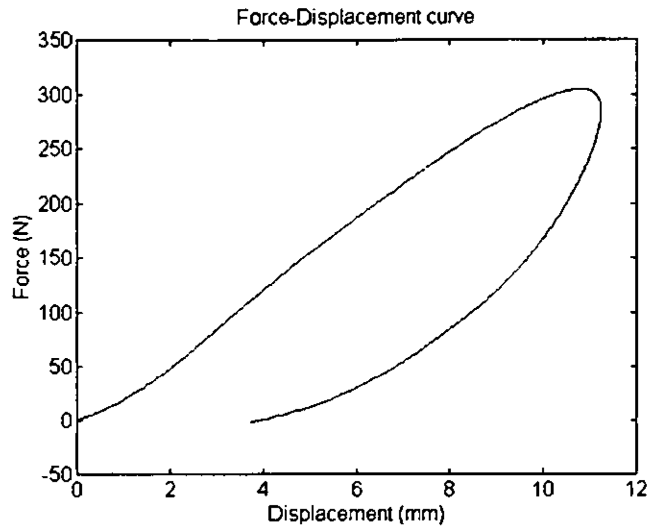


Figure 10.10- Force - displacement curve shod

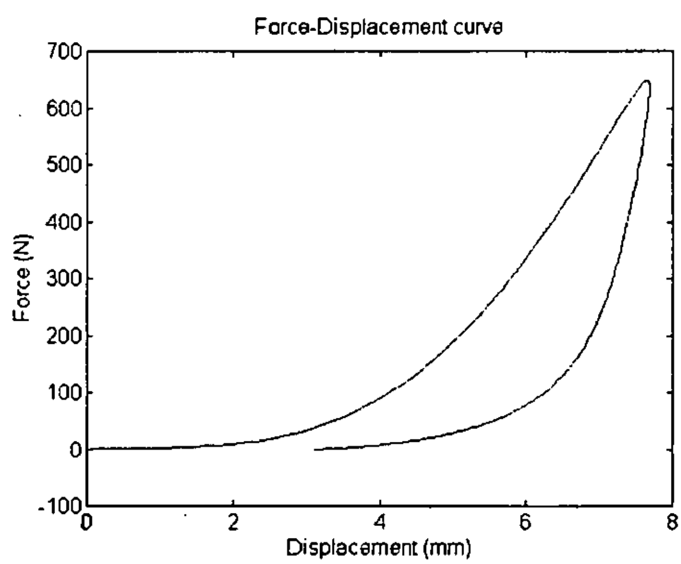


Figure 10.11. Force - displacement curve barefoot

As expected, impact force and rate of impact force were much greater barefoot with some differences between insoles (Tables 10.6 and 10.7). However, impulse (integration of force on time) was lower barefoot with some differences between insoles (Figure 10.12).

Table 10.6. Impact force (N) homogenous groups for insoles

Material	Mean Bonferroni	Groups
bb28	281.31	X
bc1	284.62	XX
bt1	285.88	XX
k1	286.75	XX
t1	292.80	XX
ab1	292.83	XX
bb21	292.93	XX
ab2	295.41	X
ab36	299.20	X
ab37	300.66	X
des	542.16	X

Table 10.7. Rate of loading (kNs^{-1}) homogenous groups for insoles

Material	Mean Bonferroni	Groups
bb28	11.10	X
bc1	11.21	X
bt1	11.22	X
k1	11.35	X
t1	11.60	XX
ab2	11.64	XXX
bb21	11.72	XXX
ab1	11.78	XXX
ab37	12.30	XX
ab36	12.38	X
des	36.37	X

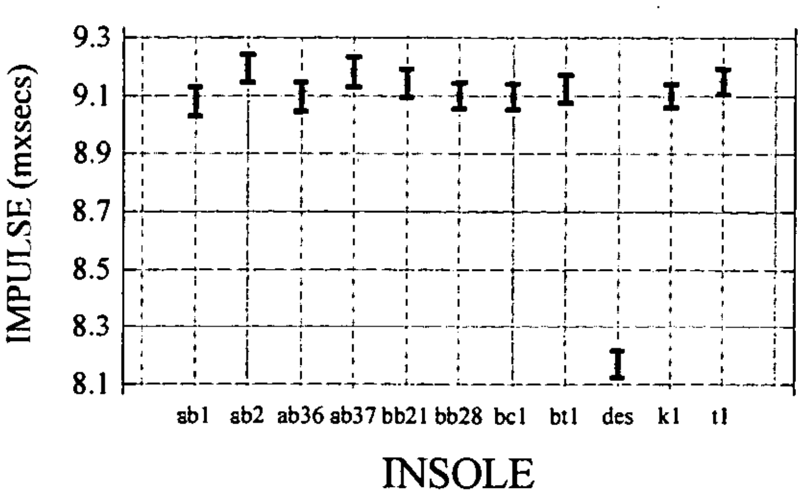


Figure 10.12. Impulse for insoles. Des = barefoot.

Wall and tibia acceleration presented differences both between participants and between insoles. These were greater barefoot (Figures 10.13 and 10.14) with, at the same time, some differences observed between insoles. Lower tibia acceleration was measured for T1 and greater for BB28 and AB1.

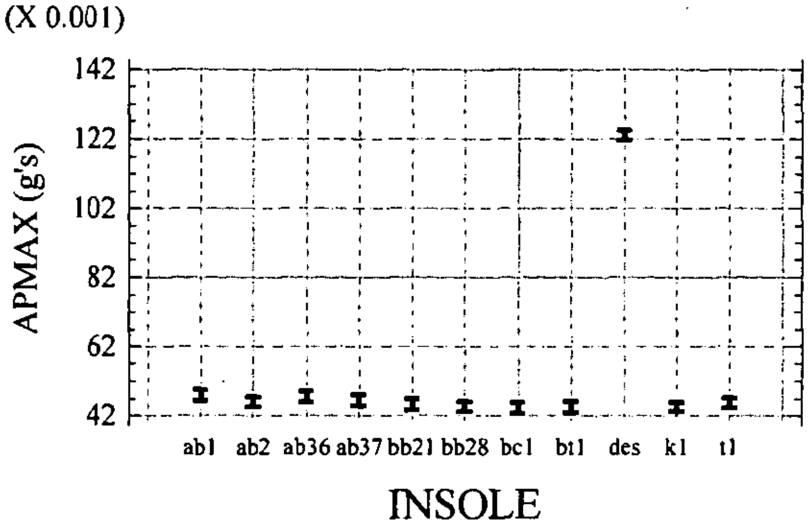


Figure 10.13. Wall acceleretion for insoles. Des = Barefoot.

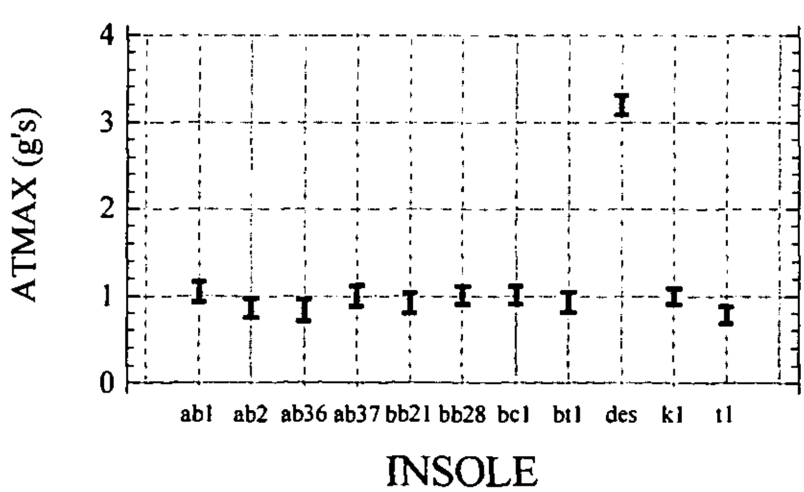


Figure 10.14. Tibial acceleretion for insoles. Des = Barefoot.

Transmission parameters showed interesting results. Transmision between impact force and wall acceleration (RAPAR) was greater barefoot followed by AB1 with no further differences between insoles (Figure 10.15). Transmission to tibia was also greater barefoot showing great differences between insoles (Figure 10.16). It was lower for T1 whilst quite homogenous groups were found among insole materials.

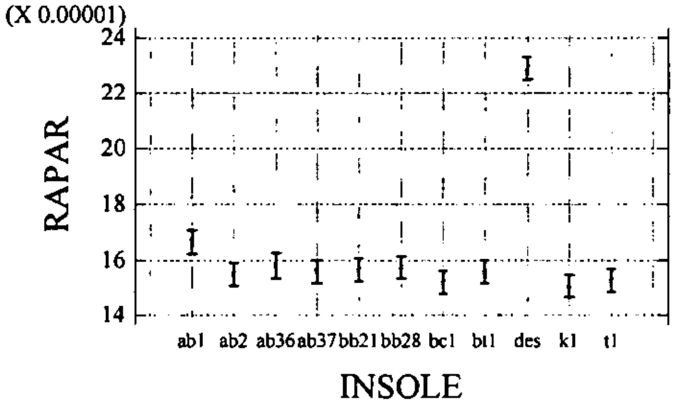


Figure 10. 15. Well transmission (Apmax/Fzi) for insoles. Des = Barefoot.

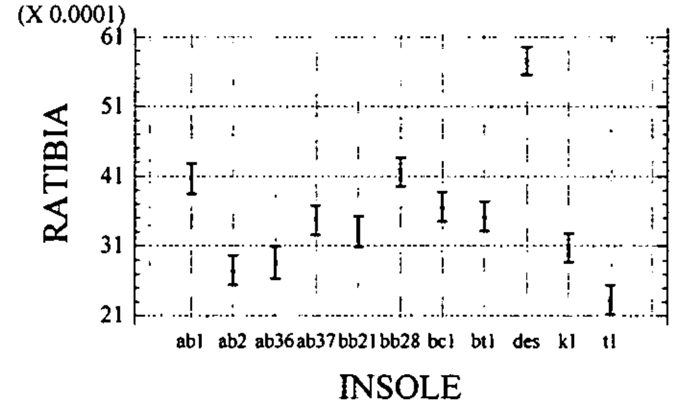


Figure 10.16. Tibial transmission (Atmex/Fzi) for insoles. Des = Barefoot.

Transmission between tibia and wall showed significant differences, but no differences between barefoot and some insoles (Figure 10.17).

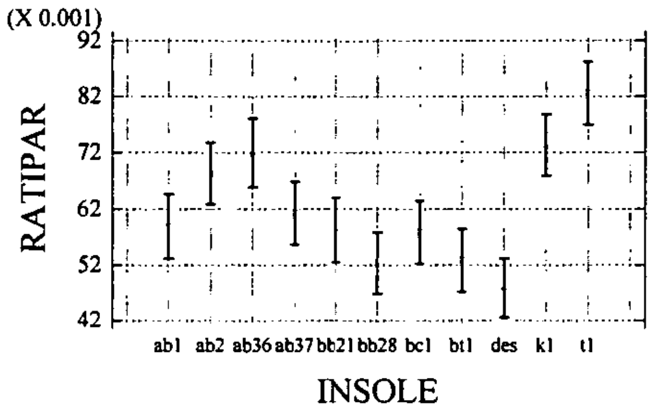


Figure 10.17. Wall/tibie acceleration transmission for insoles. Des = 8arefoot

With respect to the heel pad, peak displacement was lower barefoot and greater for K1 (Figure 10.18) whilst the energy absorption was greater barefoot, followed by K1 and lower for AB2 (Figure 10.19).

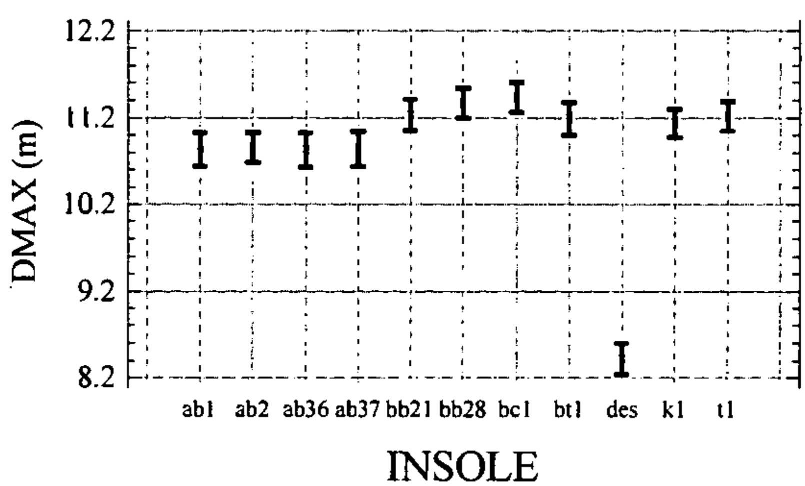


Figure 10.18. Peek displacement for insoles. Des = barefoot.

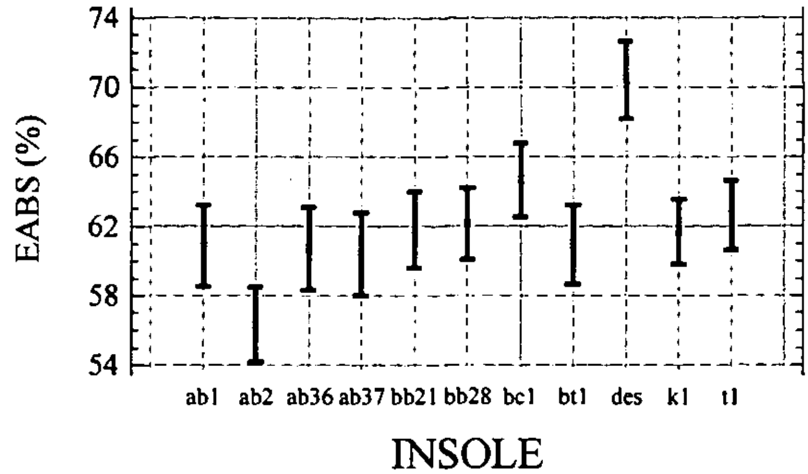


Figure 10.19. Energy ebsorption for insoles. Des = barefoot.

Stiffness parameters showed significant differences. These were much greater barefoot for all variables (Kfmax, Kini, Kmax, Kfinal), and could be due to greater and faster impact forces since the stiffness of viscoelastic materials increases as loading speed increases (the heel pad and most insoles are made of viscoelastic materials).
The residual displacement was lower for AB2 with no differences between insoles and barefoot. It showed, however, great differences between participants (Figure 10.20), higher for UV3 and lower for UV10 and UV5. Residual displacement appears to depend more on the participant.

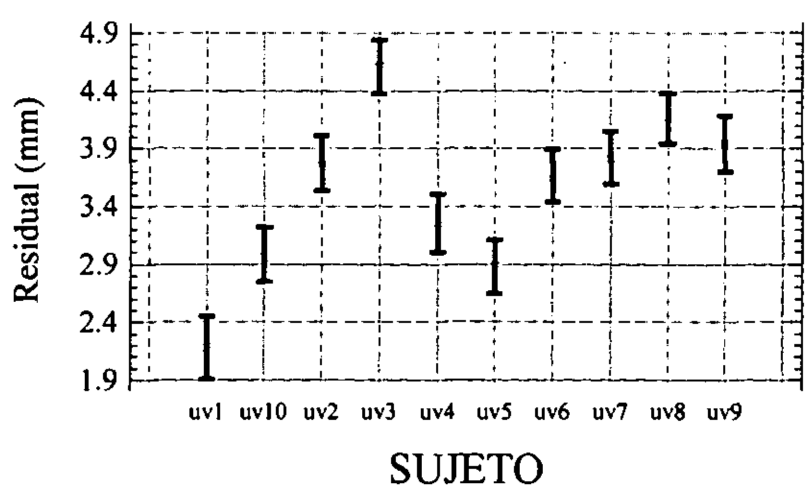


Figure 10.20. Residual displacement for participants

Shift time showed significant differences between insoles but no difference between barefoot and insoles (Figure 10.21). Only insoles AB7, AB36 and T1 showed lower time than barefoot.

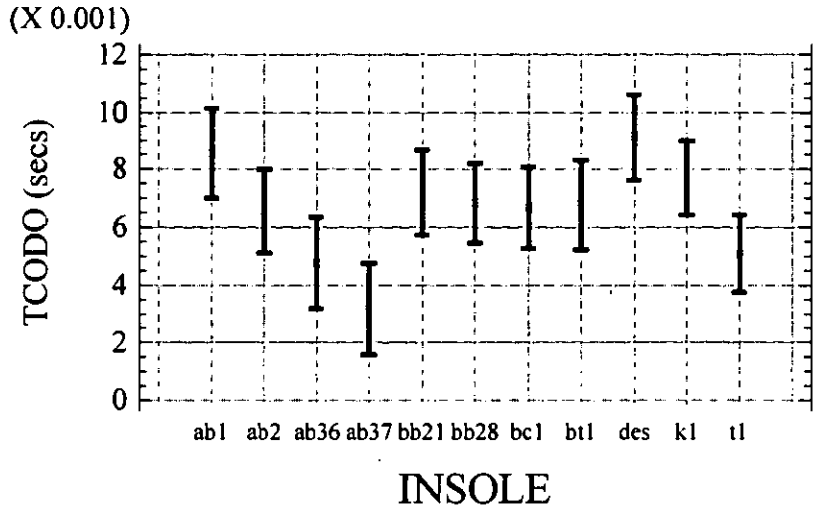


Figure 10.21. Shift time for insoles. Des = barefoot.

10.3.3. Correlation analysis

The purpose of this analysis was to study the impact mechanics and transmission properties of the lower leg both barefoot and shod. The lower leg can be considered as a mechanical system comprising shoe, insole material, heel pad and lower leg. Only principal components for all levels were studied whilst Eabs, Kmax, Kini, dmax and TCodo were considered to represent heel pad properties and peak force, rate of loading and tibia acceleration, namely the biomechanical variables at both participant and insole levels.

At all level

The results showed a high positive correlation between the elastic deformation component and the impact-wall component (0.754) and low negative correlation with tibia-rate component (-0.470) whilst the compression elastic component (initial and maximal stiffness) correlated low positive (0.494) with tibia and very low (-0.278) negative with impact. The viscoelastic component showed no significant correlation. In this sense, deformation and final compression seemed to play an important role in impact-wall parameters whilst both heel pad elastic components showed a reverse relationship with rate-tibia.

At insole level (COV=0.889)

At insole level many statistical significant correlations were observed (Table 10.8).

Table 10.8. Correlation results at insole level

	ATMAX	Dmax	Eabs	Fimpact	Kini	Kmax	RATFIM
Dmax	-.927	1.000					
Eabs	.808		1.000				
Fiimpact	.976	-.970	.743	1.000			
Kini	.947	-.911	.772	.959	1.000		
Kmax	.985	-.964	.740	.994	.962	1.000	
RATFIM	.981	-.966	.757	1.000	.961	.996	1.000
Tvisco	-.677	.803		-.735	-.683	-.751	-.725

As at all level, impact force correlated high positive with rate of loading and tibia acceleration. Tibia acceleration correlated high negative with peak displacement and delay time and positive with energy absorption and both stiffnesses. Peak force and rate of loading showed similar correlations.

Participant level (COV=0.889)

The results showed some statistically significant correlations (Tables 10.9 and 10.10). Insole peak force showed only moderate positive correlation with barefoot peak force and maximal stiffness whilst tibia acceleration only showed positive correlation with barefoot value. Rate of loading, however, correlated with both barefoot maximal stiffness and delay time, high positive and high negative respectively.

Table 10. 9. Correlation results for heel pad properties barefoot and shod at participant level.

	DMAX	Eabs	Kini	Kmax	Kinides	Dmaxdes	Eabs	Kmaxdes	TVISCO
Eabs		...							
Kini			...						
Kmax		-.373	.818	...					
Kinides					...				
Dmaxdes						...			
Eabsdes		.633				.664	...		
Kmaxdes								...	
TVISCO						-.700		.642	
TVISCODE									

Results showed that maximal stiffness barefoot accounted for great part of variance (> 60%) of impact forces and rate of loading both barefoot and shod. Thus, at participant level, barefoot properties of heel pad played an important role in determining mechanical behaviour when shod with an average insole. Barefoot properties of the heel pad could be used to explain great part of variance of what happened when shod.

Table 10. 10. Correlation results for heel pad mechanics barefoot and shod at participant level.

	RATFIM	RATFIMDES	ATMAX	ATMAXDES	Fimpact	FimpactDES
Dmax						
Eabs						
Kini						
Kmax						
Kinides						
Dmaxdes						
Eabsdes						
Kmaxdes	0.909	0.789			0.816	0.852
Tvisco	-0.686	-0.836				-0.748
Tviscodes						
RATFIM		0.741			0.933	0.835
RATFIMDES						0.725
ATMAX				0.634		
Fimpact						0.779

10.4. Discussion and conclussions

The impact forces and heel pad properties measured barefoot in this experiment were in the range of previous studies (Chapter 7). There were, however, little data on shod heel pad for comparison and even less about the effect of changing insole material.

The heel pad mechanics for both barefoot and shod are defined by the same three principal components, namely global elastic behaviour, initial deformation and viscoelastic behaviour (Chapter 7), but the components have different meaning (Table 10.11). When shod the viscoelastic components were similar but the behaviour of the elastic components was significantly different: shod global deformation appeared to be related to final stiffness and displacement whereas barefoot deformation was related to initial stiffness whilst initial behaviour shod included maximal stiffness. In general, it was observed that maximal deformation was linked to lower stiffness: initial stiffness when barefoot and final stiffness when shod. This was the main difference deduced from force - displacement curves. Thus, the heel pad shod showed a variation in properties mainly in the elastic behaviour, principally in the *initial* behaviour in a way that initial (which in turn increased) and maximal stiffness regions were linked whilst the initial of low stiffness deformation observed when barefoot was dramatically reduced. This confirms results found in the literature. De Clercq et al. (1993) described a mechanical coupling between footwear material and heel pad that increased global stiffness depending on the relative stiffness of both, with the shoe determining initial stiffness and global deformation causing deterioration of heel pad function.

As expected, Anova results highlighted greater and faster impacts and accelerations barefoot, but lower impulse levels. However, greater energy absorption and stiffness values were measured barefoot with lower peak displacement. Nonetheless, no significant differences were found between barefoot and the different insoles for shift time. All transmission parameters showed differences except the wall - tibia transmission.

The residual displacement ranged between 2 mm and 7 mm and appeared to be mainly human-dependent with no significant differences between barefoot and shod. As observed in this work and confirming Valiant (1984) the initial low stiffness barefoot deformation phase continues for the first 5 mm of deformation with 2 mm of lateral displacement for heel pad reshaping. This mechanism has been described as relevant for heel pad function. However, since there is a residual deformation in the range of the initial low stiffness deformation, initial elastic deformation will be reduced or even eliminated.

Table 10.11. Comparisson of Principal Components Analysis results from barefoot and shod

	Factor 1		Factor 2		Factor 3	
VARIABLE	Barefoot	Shod	Barefoot	Shod	Barefoot	Shod
Dmax	-0.531	-.750	0.832			
Eabs					0.958	.871
Kfinal	0.855	.870				
Kini	0.520		-0.800	.793		
Kmax	0.897			828		
Kmean	0.734	.977	-0.657			
RESIDU						.938
TCODO			0.952	.911		
Tdmax	-0.632	-.950	0.740			
TVISCO				.594	0.918	-.762
TDURA	-0.649		0.646			
Eigenvalue After rotation	3.826	3.446	3.541	2.868	1.941	2.647
% Explained Variance	38.26	34.46	35.41	28.68	19.41	26.47
% Cumulative Variance	38.26	34.46	73.67	63.11	93.08	89.65

The residual displacement ranged between 2 mm and 7 mm and appeared to be mainly human dependent with no significant differences between barefoot and shod. As observed in this work and confirming Valiant (1984) the initial low stiffness deformation phase barefoot continues for the first 5 mm of deformation with 2 mm of lateral displacement for heel pad reshaping. This mechanism has been described as relevant for heel pad function. However, since there is a residual deformation in the range of the initial low stiffness deformation, initial elastic deformation will be reduced or even eliminated.

Nevertheless, differences were found between insoles for many variables demonstrating that it was not only shoe effect what influenced results and that insoles played an important role in passive shock absorption due either to the materials properties or to passive interaction.

Principal Component analysis for impact and transmission parameters indicated that two components could explain the 88.8 % of total variance. The first - which explained near the 60 % of variance - was determined by impact and wall acceleration related parameters, both absolute and normalised, whilst the second described a relationship between tibia acceleration and impact rate of loading, again absolute and normalised. Thus there are two components, one describing higher/lower impacts and wall accelerations and the second describing faster/slower impacts and higher/lower tibia accelerations.

The correlation analysis between principal components indicated that for the shod heel pad, elastic deformation component could account for most of the variance of impact forces and wall acceleration in such a way that both decrease with decreasing component which means increasing deformation. Lower negative relationship was observed with loading rate and tibia acceleration, thus they decrease with decreasing deformation. The elastic loading (initial-maximal component) behaviour acted in a positive sense over the latter: that is, stiffer heel pad with longer initial low stiffness region related to greater tibia acceleration and faster impacts. The viscoelastic component showed no correlation.

Analysis at participant level showed that barefoot maximal stiffness played an important role on shod heel pad mechanics in terms of explaining variance of impact parameters whilst at insole level it was observed that elastic (peak displacement and stiffness) and viscoelastic (delay time and energy absorption) properties had a greater influence on heel pad mechanics explaining part of the variance of tibia acceleration, peak force and rate of loading. Thus for a person wearing an average insole (insole variability eliminated to assess human effect) it is barefoot global stiffness that determines the impact whilst for an insole worn by an average person (human variability eliminated) impact is described by the combination of elastic and viscoelastic properties.

When the foot is on the ground, the interface comprises three basic features, namely the shoe-insole materials, the heel pad and passive interaction. Principal components and correlation analysis indicated that both passive impact forces and wall acceleration decrease with increasing deformation of the system (i.e. lower elastic deformation component) which increases tibia acceleration and rate of impact loading. At the same time, the tibia acceleration and rate of loading increase appear to be related to increase of initial-maximal stiffness component of the system. Thus, to a great extent, mechanical low stiffness deformation determines the global mechanical response. This will depend on the relative stiffness and viscoelasticity of the heel pad and insole material. As reflected by results at participant level and insole level the influence of heel pad on system's elastic behaviour is determined by its maximal stiffness whereas the influence of the insole depends on different elastic and viscoelastic parameters.

On the other hand, there is a residual displacement that reduces or eliminates the initial low stiffness area of the heel pad when shod. As a consequence, the heel pad contributes mainly to initial-maximal stiffness and to viscoelastic components whereas shoe-insole will influence all components. The elastic deformation behaviour linked to final stiffness could be attributed to shoe confinement as reported by De Clercq et al. (1990).

Figure 10.22 demonstrates the influence of relative softness in passive interaction. If the insole was stiffer than the heel pad then at initial loading mechanical coupling, this leads to increased contact area thus modifying plantar pressures causing high residual deformation of the heel pad (probably due to bottoming out) which, in turn, changes the properties of the heel pad. In contrast, if the insole material were softer than the heel pad, residual deformation of insole would confine the heel pad and modify heel pad properties too. In general, similar heel pad and insole material is preferred for optimum heel pad and increase in contact area. However, this may be difficult in practice as both heel pad and insole materials have viscoelastic properties and, therefore, stiffness of the system depends on rate of impact loading. To understand this phenomenon better, an analysis in frequency domain is needed for an increased understanding of shock transmission. At the same time as loading frequency changes the stiffness of the system, passive interaction could also be modified. If the heel pad was more frequency dependent than insole any increase in frequency will stiffen the heel pad relative to the insole material and, in turn, modify passive interaction. All these aspects are of high interest for future work.

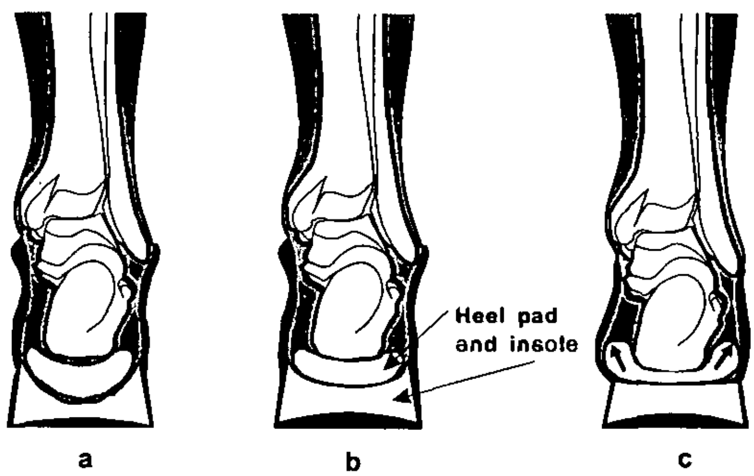


Figure 10.22. Influence of relative stiffness in passive interaction: a) insole softer than heel pad, b) insole stiffness similar to heel pad and c) insole stiffer than heel pad

In conclusion, elastic deformation of the system increases to reduce impact forces mainly by shoe-insole deformation, which, as a consequence, increases tibia acceleration and rate of loading. But reducing initial-maximal stiffness component - achievable through good heel pad-insole interface design - could compensate the latter effect. This would require a good knowledge of heel pad properties to select appropriate insole material.

11

Analysis and conclusions.

Contrasting hypotheses and answering questions

11. Analysis and conclusions

11. Analysis and Conclusions

11.1. Introduction

The aim of this research was to ascertain the most satisfactory of the statements proposed to explain the differences between walking and machine testing of shock absorbing materials. The work described in previous chapters attempted to answer some of the questions that arose in relation to these statements. In the course of that work data was collected and partial results deduced and described. These partial results are analysed in this Chapter with respect to the different research issues established were as follows:

1. **That there is no accurate simulation of impact loading by machine test.** This statement was investigated initially by comparing a machine test simulating impact loading with walking tests.
2. **That materials degrade in use showing lower properties than measured by machine testing.** A new testing method was developed to characterise the ability of materials to recover after compression and the results compared with human testing results.
3. **That shoe effect on walking kinematics and heel pad confinement have greater influence on shock absorption than the properties of underfoot materials.** An instrumented pendulum was developed to analyse heel pad mechanics and its dependence on human physical characteristics. In further experiment using the pendulum test, heel pad mechanics and shock absorption data obtained under passive conditions using barefoot participants and shod participants with the same shoe type and different insole materials. Results were analysed together with regard to material properties and shock absorption, pressure distribution and rearfoot movement during walking.
4. **That human accommodation to impact conditions occurs according to a controlled proprioceptive feedback model developed by Robbins (1989-1991).** This research issue raised many questions about the shock absorption theories that are currently under investigation. An important number of these questions have been studied in course of this research by correlation analysis of objective, subjective and material properties.

11.2. Statement 1. There is no accurate simulation of impact loading in machine test

The hypothesis proposed was that testing materials simulating the magnitude and duration of walking impact forces would yield a good prediction of material's performance during walking as given by ground reaction forces and accelerations.

The work carried out to test this hypothesis included the use of a machine testing method simulating impact loading and comparison of results with walking tests. The work and results have been described in detail in Chapter 6 and published in the Journal Foot and Ankle International (Forner at al., 1995). The machine method developed at IBV (Institute of Biomechanics of Valencia) by García et al. (1994) to simulate impact loading history with respect to impact force and tibia shock reduction during walking showed good correlation with human testing. However, accommodation was considered to be the likely cause of the discrepancies observed in the results between the tibia and the head.

It was concluded that, according to the proposed hypothesis, accurate simulation of loading history improves the ability of machine testing to predict walking performance of materials in terms of force and acceleration. This is the case for results between the foot and knee. However, between the knee and the head accommodation occurs, the extent depending on underfoot stiffness, and this changes loading history and this phenomenon was further investigated as described in Chapter 9. That investigation, however, failed to find any correlation of machine testing parameters with impact forces and accelerations. This failure was attributed to the influence of passive interaction on heel pad mechanics and tibia transmission that became evident from pendulum tests with shod people, Passive interaction (mechanical coupling between heel pad and insole) was studied initially by placing a material insert only under the heel (Chapter 6) whereas subsequently a whole insole was inserted. In the latter case, the loss tangent of materials showed lower values and this may account for the nil correlation. However this implies that loss tangent (i.e. the viscoelastic component) played a very important role which was not supported by further results in which viscoelastic properties were found to play a small role.

Thus, the results obtained in Chapter 6 apply to heel insert materials. In this sense, in general materials with a high loss tangent and moderate rigidity should be preferred for reducing impacts transmitted from ground to the tibia and forehead. Optimum rigidity would yield diminished transmission to tibia without lowering impact proprioception and, therefore, not increasing the transmission of impact from tibia to forehead. These results would only apply for people with a normal proprioception and without impaired shock attenuation systems. For the elderly and others with impaired shock attenuation systems less rigid materials are preferred, as these people cannot adapt to perceived high impacts. Besides, for some patients (neuropathic diabetes, metatarsalgia, etc.) it may be more important to avoid peaks of plantar pressure by means of a less rigid material. Nevertheless, further research in accommodation was necessary and described below (11.5.).

11.2.1. Statement 1 - Conclusions

The results of this research have demonstrated that accurate simulation of loading history improves the ability of material testing to predict walking performance of insert materials – up to the knee - in terms of force and acceleration. Between the knee and the head accommodation occurs and this introduces differences depending in underfoot stiffness. However, when a complete insole was considered the good agreement between machine and human tests was not evident. This was attributed to passive interaction between insole and heel pad and this requires further investigation.
Thus, in general, heel insert materials with a high loss tangent and a not very low rigidity should be preferred for reducing impacts transmitted from ground to the tibia and forehead. Optimum rigidity would yield diminished transmission to tibia, without lowering impact proprioception and, therefore, without increasing the transmission of impact from tibia to forehead. For insole materials, the machine test used did not produce a good prediction of walking results so human testing became necessary.
Thus, the work conducted so far has demonstrated that Statement 1 is not true and accommodation was suggested as the reason for discrepancies between machine and human testing.

11.3. Statement 2. That materials degrade in use showing lower properties than measured by machine testing.

Bottoming out occurs when the material under load is compressed beyond its effective thickness with subsequent compressive collapse and loss of properties. This statement implies that materials collapse in use and do not recover between steps with an accumulative loss of properties. Thus, walking properties are different than those resulting from machine testing. To explore this statement two hypotheses were proposed:

1. That the recovery-ability of materials related to the residual compressive displacement measured using a machine test simulating walking underfoot pressures is related to long term loss of performance of materials when walking as evidenced by reduced levels of subjective perception.
2. Those materials with higher residual displacement cause greater and quicker impact forces and higher accelerations during walking.

These hypotheses were tested by comparing results from walking tests with those from a new testing method (developed as part of this research) that measured the recovery-ability of materials by simulating foot plant pressures. Testing bottoming-out in use require *in vivo* measuring of heel pad and insole compression and could be achieved only by expensive and complex techniques such as x-rays filming or cyneradiography (that may not accurate enough) (De Clercq et al., 1990). For this reason a different approach was chosen. The aim was to determine the ability of materials to recover under compressive loading and to assess whether this was related to differences observed between human and machine testing. A new machine method was developed and a selection of materials was tested and the results compared with those from walking tests. No similar parameter or test were found in literature although some reports refer to energy return meaning impact energy that was either dissipated or returned to the foot (Nigg & Herzog, 1994). Results from machine testing on different insole materials were compared to results from subjective testing to check the first hypothesis and with walking tests to assess the second hypothesis.

11.3.1. Material and Methods

The dynamic cushioning test (Chapter 8), to simulate plantar pressure loading was used to measure the recovery-ability of materials. Several parameters (compressive work, residual deformation, etc.) that describe this ability to recover after removing a compressive load were obtained from the force and displacement signals.

One parameter (Rebote) measured the residual displacement of materials when load reached the zero magnitude signalling structural collapse and loss of mechanical integrity. It was obtained from each of the ten loading sets used in the cushioning test (Chapter 8), namely, pressure of 100, 200, 300, 400 and 500 kPa applied for 400 and 800 milliseconds. But, after variable reduction analysis using hierarchical clustering statistical analysis, this parameter was studied only for loading sets 28 (Rebote28) and 44 (Rebote44) which corresponded with pressures occurring under the 5th metatarsal head and under the heel and metatarsal heads' areas. This reduction allows the variance of recovery at all the other loading histories could be considered by reference to these two sets only - one of high load and high velocity (44, 400 kPa for 400 milliseconds) and the other of low load and low velocity (28, 200 kPa for 800 milliseconds).

Pearson's correlation analysis was used to study the influence of these parameters in shock absorption in walking tests. Correlation analysis was also done with plantar pressure distribution, rearfoot movement and results from pendulum testing. Correlation was also done to test the relationship of Rebote with long term performance of materials, with subjective opinions using the Kruskal-Wallis mean rank of subjective opinions as a measurement of central tendency. These analyses at insole level reflect the response of participants in these experiments when wearing the insoles.

11.3.2. Results and discussion

In machine testing for loads applied dynamically and statically, no material showed either a peak displacement or residual displacement approaching its original thickness. The insoles were 6 mm thick whilst peak displacement ranged between 0.08 mm and 2 mm whilst residual displacement ranged between 0 mm and 0.29 mm. Thus bottoming out did not occur during machine testing. Nevertheless, the recovery-ability analysed in this Chapter did not really measure bottoming out as material collapse, but reflected the integrity of material from step to step.

Regarding the first proposed hypothesis to explore this statement, Rebote44, which resulted from loads similar to under-heel pressures, showed a rather logical high positive correlation (0.728) with heel material perception, the greater the residual displacement the harder the heel was perceived to be. At the same time, Rebote28, for a low and slow loading which simulates pressures under the 5th metatarsal heads, correlated high negative ($r = -0.748$) with impact perception in the walking trial experiment which reflected a negative linear relationship of increasing residual displacement and perceived harder impacts. Both subjective opinions were a result from long trials (1 hour) in which the bottoming out may become more important as the material loses integrity. This result could indicate that there exists a certain relationship between the recover-ability of materials and their long-term performance and that the study of recovery-ability requires a long-term analysis.

These results are important for footwear design since greater residual displacement for pressures taking place under the 5th metatarsal heads is related to perception of harder impacts leading to accommodation, which is generally undesirable (Chapter 4). These results are also important for this research with respect to the Robbins model, On the other hand, concerning the second hypothesis the results obtained from correlation analysis with walking tests showed, in general, that Rebote28 and 44 have little influence in shock absorption, pressure distribution and rearfoot movement. Nevertheless, Rebote28 showed a positive moderate correlation with rate of impact force loading which could be due either to (i) direct loss of material thickness or stiffness increase, both of which could explain the force rate increase, but without collapse since no correlation with either impact forces or accelerations were found or (ii) accommodation due to increased impact perception. This may be related to passive interaction since the greater the residual compression, the deeper the heel pad interaction with the insole either by confinement or by surface contact increase. However, no correlation was found for Rebote28 with passive properties measured in the pendulum test. This would require further research.

11.3.3. Statement 2 - Conclusions

The recovery-ability of materials measured with the new method developed for this research showed a relationship with long-term performance of materials. Rebote28 (low and slow load) accounted for the active long term impact perception (<50%) which was an important result for participants' accommodation whereas residual compression for a high and fast load (Rebote44) accounted for around 50% of variance of heel material perception. Materials showing a greater residual compression yielded perception of higher impacts and of harder heel material.

By contrast, the recovery-ability parameters showed little relationship with the variance of results obtained for shock absorption, pressure distribution and rearfoot movement. Only structural loss of mechanical integrity in terms of residual compression for a low and slow load (Rebote28) accounted for part of the variance of rate of loading when walking (<40%). These results suggest that as residual displacement increases and material thickness decreases there is a stiffness increase which could explain the increase of rate of loading as well as impact and material perception but without collapse since no correlation with either impact forces or accelerations were found.

These results established a further relationship between underfoot material and accommodation that was later thought to modify the proprioceptive model. This again would require further research involving long term trials. The recovery-ability could also be related to passive interaction since the greater the residual compression the deeper the heel pad is confined into or compressed against the insole. The recovery-ability of materials was not able to explain the differences between machine and walking testing. Nonetheless, further work needed to study the long-term properties of materials.

11.4. Statement 3. The effect of the shoe in walking kinematics and heel pad confinement has greater influence on shock absorption than underfoot materials

This statement argues that the effect of shoe on natural shock absorbing systems, such as walking kinematics and heel pad confinement, has greater influence on shock absorption than underfoot materials in walking tests that machine testing failed to predict. A specific study would be required to quantify the effect of both heel pad confinement in shock absorption and the relative effect of rear shoe design with respect to material properties. Different shoe constructions should be carefully designed and tested in walking experiments with different insole materials, as it is difficult to isolate absolute geometrical influence from possible complex accommodation leading to kinematic adjustments. Besides, the literature on heel pad confinement is contradictory reporting both improvement and worsening of heel pad properties. The definition of the amount of restraint to be provided by confinement is also unclear (Chapter 7) and previous work at IBV found that the relationship between heel pad confinement and foot movement in the shoe is dependent on many factors such as lacing, arch support, insole pitch and covering material. Thus, the analysis of heel pad confinement and kinematics would require a complex and specific study outside the scope of this study. Nonetheless, a deeper understanding of heel pad properties was needed to examine passive interaction and the associated role of insole materials. For this purpose, pendulum testing of barefoot and shod humans became part of this research but, since the aim was to compare the influence of shoes and insole materials, a different approach to testing was adopted.

In this research, the same shoe was tested with ten different insoles, which varied only in the type of insole material. Therefore, for all insole types, shoe geometry and shoe materials and heel pad confinement are common so the shock absorption characteristics are constant. Thus, any difference between insoles would be due to the material (a mechanical process) or accommodation (passive interaction) rather than the shoe.

In pendulum testing accommodation is absent so only the mechanical contribution and passive interaction could provoke differences between insoles. *Passive interaction* has been defined in this work as the resulting underfoot properties due to initial elastic deformation and stiffness modification by heel pad and insole mechanical interaction. Depending in the relative stiffness of both heel pad and insole material and the rate of applied loading, initial reshaping of heel pad will be constrained in the insole by

different heel pad confinement or by compression against the insole with increased the contact area. This will modify shock absorption characteristics as well as accommodation that have been previously related to underfoot stiffness (Chapter 6).
The hypothesis proposed in relation to this statement was that any differences found in impact forces and accelerations registered during pendulum testing on participants wearing the same shoe type but with different insoles are due insole materials and their influence on shock absorption or passive interaction. If these differences arose only during walking tests they were attributed to the influence of insole materials on accommodation. The null hypothesis was considered to support this statement as the source of differences.
To test this hypothesis, pendulum and walking tests and materials testing were carried out and various analyses completed as described below.

11.4.1. Material and Methods

There were ten participants and each tested ten different insole materials in the same shoe. Different tests were conducted to assess the role of both passive interaction and materials' properties in shock absorption. The goal was to analyse the relative influence of insole materials and shoe in shock absorption. This work was conducted as follows:

1. Barefoot heel pad impact mechanics and their dependence on individual characteristics were studied using an instrumented pendulum. This was necessary to better understand the influence of shoes in heel pad mechanics and passive interaction. It has been described in detail in Chapter 7 and sent for publication to an international journal.
2. Results from walking and pendulums tests on people wearing different insoles in the same shoe type were analysed. Differences in walking tests between barefoot and shod were considered as the shoe effect but differences between insoles were considered as confirmation of a materials effect No differences between insole indicate that materials had no influence on the outcome of the tests.

However, the influence of the materials was thought to be a combination of inherent shock absorbing properties, a materials component in accommodation and passive interaction (i.e. resulting properties of heel pad). For these reasons, shock-absorbing properties of barefoot and shod humans were studied using the ballistic pendulum. Since only the pendulum moves, any difference between insoles was considered to be due either to the insole materials or to passive interaction. However, if there were no differences between insoles, this would support the argument that any differences found in walking tests would be due to influence of materials on accommodation. This work has been described in chapter 10.

In both walking and pendulums tests, Analysis of Variance (Anova) was done considering insole and participant as factors. Normality and Homogeneity tests were done as described previously in this report. Multiple range Bonferroni post hoc test was done to study differences between homogenous groups. Principal components of the heel pad barefoot and shod were compared to study differences in heel pad mechanics. A correlation analysis between passive impact principal components (impact forces and acceleration) and shoe-heel pad variables at insole and participant level was done to study the influence in impact mechanics.

Finally, for those differences were found in pendulum test, it was difficult to isolate the materials' contribution from passive interaction as the latter depends on the relative mechanical properties of heel pad and insole. This was done by analysis at insole and participant level in order to eliminate the variability of participants and insole materials respectively. The correlation analysis at insole level was done with heel pad properties since it was not possible to compute principal components due to low power of test (n = 10).

Pearson's correlation analysis was done at two levels and results compared. Correlations between passive heel pad properties and materials properties obtained from machine testing were studied to evaluate the contribution of the material's properties to passive shock absorption properties. Correlation analysis of insole material's properties with walking results was also done. The comparison of both analyses was done to assess the contribution of materials. If the influence of insole material was evident in both cases, the contribution of the materials as a shock absorber was considered significant.

11.4.2. Results and discussion

Barefoot heel pad mechanics and the influence of individual characteristics are reported in Chapter 7. Results from walking tests as reported in Chapters 6 and 9 considered differences between insoles and supported the importance of insole materials with respect to the shoe. Moreover, pendulum tests, in general, reflected differences between shod and barefoot identifying the shoe effect. However, these differences were not evident for all study variables and further differences were found between insoles. In some cases, differences appeared between insoles but not between insoles and barefoot. It was concluded, therefore, that the insole played a relevant role along with shoe. Thus, support was not found for statement 3: The effect of the shoe in walking kinematics and heel pad confinement has greater influence on shock absorption than underfoot materials. Very interesting results were found in this part of the study and these are discussed below.

11.4.2.1. Heel pad mechanics

The impact mechanics of barefoot heel pad and heel pad-shoe-insole system were significantly different in elastic behaviour showing the importance of the material - heel pad interaction (mechanical coupling). For barefoot and shod, the viscoleastic behaviour was however similar. The force - displacement curves showed an initial elastic deformation at low stiffness for the barefoot heel pad which has been related to lateral reshaping and heel pad confinement (Valiant, 1984) but not for shod heel pad as did De Clercq et al. (1990) who showed similar curves by x-ray filming. For both conditions, the global elastic behaviour was more or less similar; but when shod elastic deformation was related to final stiffness whereas barefoot it was related to initial stiffness. In both cases, these are the lowest stiffness; thus both barefoot and shod heel pad elastic deformation at low stiffness is a component of a mechanical system.
In this context, the Anova results reflected, as expected, that barefoot gave rise to greater and quicker impacts and accelerations, but lower impulse loads. However, greater energy absorption and stiffness with lower peak displacement were measured barefoot. By contrast, no significant differences were found between barefoot and different insoles for shift time probably due to a great dispersion and to difficulties in identifying this parameter when shod because there was distinct discontinuity in the force - displacement curves. All transmission parameters showed differences except wall - tibia transmission.
Principal component analysis for shod impact and transmission parameters reflected two components that could explain the 88.8 % of total variance: one component related to impact and wall acceleration parameters and the other to tibia acceleration and the rate of impact loading. Impact mechanics was also different between barefoot and shod. Impact forces barefoot were reduced by decreased stiffness and viscoelastic dissipation whilst peak time was governed by the initial deformation component at low stiffness. However, shod impact forces and wall acceleration decrease with increasing deformation but greater tibia acceleration and quicker impacts will occur. At the same time, the elastic loading (initial-maximal stiffness) behaviour acted in a such a way that stiffer shod heel pad related to greater tibia

acceleration and quicker impacts whilst viscoelastisic component showed no correlation.

Analysis at participant level (when the insole variability was eliminated) showed that barefoot maximal stiffness (stiffness was shown to depend on individual characteristics (Chapter 7)) played an important role on shod heel pad mechanics in terms of impact parameters. On the other hand, at insole level (and participants of average patterns of behaviour) the roles of elastic properties (peak displacement and stiffness) and viscoelastic properties (delay time and energy absorption) in impact mechanics are enhanced.

Nevertheless, in terms of interaction, when shod an elastic deformation behaviour linked to final stiffness (deformation principal component), corresponding mainly to the shoe-insole system showed a reduced peak force and wall acceleration by increasing deformation, probably due to the shoe, as described by De Clercq et al. (1990), and as reflected by the disappearance of the initial low stiffness when shod but observed barefoot. However, quicker impacts and greater tibia acceleration occurred with greater deformation and with stiffer initial-maximal stiffness component, depending on the heel pad properties.

From these results, passive interaction between underfoot materials and heel pad that determine initial elastic deformation and following stiffness under loading should be defined as the mechanical coupling between both. This will depend on the relative stiffness and the dependence of underfoot materials and heel pad properties ~~in~~ on loading frequency (and the viscoelasticity effect). This would be supported by the fact that residual displacement is in the range of the observed low stiffness displacement (Valiant, 1984). The residual displacement ranged between 2 mm and 7 mm and appeared to be mainly participant dependant. That represents a maximum residual displacement of about 30% of heel pad thickness (20 mm) whereas for insole it was of 5% (3.3 mm). This residual compression will reduce and even eliminate the initial low stiffness deformation phase which occurs over the first 5 mm of deformation (Valiant, 1984). The distribution of this displacement between the heel pad and the insole remains to be clarified and will, probably, determine the influence of passive interaction. This passive interaction depends on relative stiffness of insole and heel pad (See Figure 10.22). Greater deformation of insole material would increase heel pad confinement whereas greater heel pad deformation would lead to a greater contact area between heel pad and insole thus reducing pressures. The change in pressures - highlighted in this thesis and observed by Robbins et al. (1988) - will ultimately influence perception of forces and thus influence accommodation. This is one of the most relevant results of this research and worthy of further work.

Increased deformation with reducing forces is mainly achieved by the combined contributions of insole and shoe (Chapter 10). The heel pad loses the initial reshaping - due to confinement into the insole or to increased contact area due to compression (Chapter 7) - and also its initial low elasticity and thus increasing its initial-maximal stiffness which in turn increases the shock transmitted to the tibia linked to quicker impacts, probably with a different frequency content than that reaching the wall. This depends on relative insole – heel pad stiffness. However, the heel pad is more viscoelastic and, therefore, more sensitive to differences in loading rate than the insole and thus, differences in loading frequency will influence mechanical coupling.

This is supported by results of this research that show that the greater the residual compression the greater the rate of loading during walking and related to Statement 2. This should be further investigated in the frequency domain. In reports of human pendulum tests comparing kinematics response with heel pad contribution for different interface materials Lafortune et al. (1995a) concluded that the combined stiffness of heel pad and material interface totally govern the rate of application of force during initial impact. This supports the above results.

11.4.2.2. Correlation analysis between passive testing and material properties

The statistically significant correlations are presented in Table 11.1. Parameters from dynamic impact testing showed no significant correlation (COV = 0.889) with passive properties. Hardness showed negative correlation with peak displacement of heel pad, which also correlated negatively with density and global stiffness (RT) and positively with different elastic and viscoelastic parameters from dynamic cushioning (Tvisco, Tcodo and Tdmax).

Energy absorption and delay time (viscoelastic parameters of shod heel pad) correlated negatively with density. Energy absorption showed a negative correlation with some elastic properties of materials such as global stiffness (RT), Kmax38 and tcodo58. Elastic properties of heel pad (Kmax) showed a positive correlation with density and a negative correlation with tcodo42 and tdmax18 whilst initial stiffness (Kini) correlated positively with similar parameters as tcodo41 and tdmax18.

Table 11.1. Correlation passive - meterial properties

	Atmax	Fimpact	Ratfim	Dmax	Eabs	Kini	Kmax	Tvisco
Density				-0.695	-0.672		0.731	-0.644
Hardness		.816	.800	-0.723				
RT		.704	.672	-0.741	-0.681			
D300		-.698						
D250		-.731	-.654					
Eabs18	.654							
Eabs24		-.678						
Tvisco42								0.650
Tvisco44			-.705	0.760				
Tvisco45		-.883	-.780	0.670				
Tvisco48	-.710	.632						
Kini18	-.736							
Tcodo18	.659							
Tcodo41						0.739		
Tcodo42				0.692			-0.876	
Tcodo58					-0.655			-0.659
Kmax18	-.639							
Kmax38					-0.706			-0.675
Kmax41	-.639							
Kmax58		-.668						
Tdmax18		-.688	-.733	0.873			-0.688	
Tdmax28						0.668		
Tdmax41		-.751	-.745	0.734				
Tdmax43	-.733							
Tdmax44	-.721							
Tdmax48	-.717							

Impact parameters, peak force and rate of loading, correlated highly positive with hardness, and moderately negative with elastic parameters such as displacement (D300 and D250) from static compression, maximal stiffness (Kmax58, r = -0.668), time-to-peak displacement (tdmax18 and 38) and viscoelastic parameters such as energy absorption at 24 (Eabs24), Tvisco 44 and Tvisco45 (-0.883) and moderately positive with Tvisco48.

Tibia acceleration correlated highly negative with initial elastic behaviour (Kini18, -0.736; tcodo42) and elastic parameters as time-to-peak displacement tdmax43, 44 and 48; maximal stiffness (Kmax18, Kmax41) and some viscoelastic parameters (tvisco48), whilst moderately positive with global static compression stiffness (RT), eabs18, and tcodo18.

Thus the materials' role in passive properties could be explained from machine testing results to some extent. Hardness and elastic material properties showed high correlation with impact parameters and elastic parameters of shod heel pad. However, quicker impacts and greater tibia acceleration occurred for greater deformation and for stiffer initial-maximal stiffness which could depend on the heel pad characteristics and passive interaction given the relevant role observed for barefoot heel pad in shod impacts and the few significant correlations found between maximal and initial shod stiffness and material properties. Besides, those found were mainly with shift time (Tcodo14, 24) and time-to-peak displacement (Tdmax18, 28) which define insole behaviour at initial loading under low and slow loads (1 and 2, respectively as described in Chapter 8).

Therefore, the elastic deformation behaviour linked to final stiffness could be attributed to the shoe as described by correlation analysis with materials, in agreement with De Clercq et al. (1990). On the other hand, hardness is a measure of scratch and indentation resistance of materials. The mode of deformation under an indentor is a mixture of tension, shear and compression so it could be related to the mechanical coupling between insole and heel pad as related to how the heel pad penetrates the insole.

These results seemed to demonstrate an important materials contribution to impact and acceleration at two levels: increasing hardness due to deformation and change in elasticity modifying the final compression part of the force-displacement curve to decrease peak forces and accelerations increasing at the same time tibia acceleration and the rate of loading. However, the elastic behaviour was more participant dependent, being modified instinctively by passive interaction at initial loading with some materials contribution supported by correlation with different material parameters that describe initial behaviour which also correlated with tibia acceleration. Increase in stiffness further increased tibia acceleration and rate of loading. In this sense, mechanical coupling between heel pad and footwear could depend on their different viscoelastic properties. Since viscoelasticity is sensitive to loading rates, the heel pad and footwear respond differently resulting in different stiffnesses which, in turn, determine the nature of the passive interaction. In this sense, at materials' level the behaviour under initial loading appeared to be important as well as hardness and final stiffness. It follows that covering and multi-layer materials may attract the interest of biomechanicists in the near future.

The viscoelastic component would be the result of human heel and insole combination, supported by residual displacement dependence in human beings. The fact that no significant differences were found for energy absorption and delay time between insoles as well as the negative correlation between energy absorption of insoles and between impact forces and accelerations suggest that the viscoelastic component has little influence on impact mechanics. Some correlation of material properties with viscoelastic passive parameters was found which suggested that passive interaction also modified viscoelastic properties of the system due either to heel pad confinement or the influence of loading frequency. Frequency analysis of heel pad properties with different combination of insole and heel pad stiffness and viscoelasticity would be required to underpin these results.

11.4.2.3. Comparison of material and passive properties correlation with active and passive shock absorption

Correlation analysis between heel pad principal components and active shock absorption showed that components from pendulum tests that included impact parameters and wall acceleration, correlated positively with walking impacts components showing a direct relationship between passive and active forces. Besides, the components from pendulum test included rate of loading and tibia acceleration - previously related to passive interaction - appeared to be related also to the transmission component during walking. Thus the material contribution to increase deformation would act on active impact parameters whilst heel pad and passive interaction would affect transmission in relation to rate of loading and tibia transmission. Furthermore, the elastic deformation component of the heel pad - most influenced by material properties - showed only a low role in active impacts whilst the initial and maximal stiffness component which is more human dependent (i.e. passive interaction) showed a negative relationship with reduction of impact parameters and acceleration with respect to barefoot (stiffer participants presented greater reduction with respect to barefoot) and a positive relationship with transmission (stiffer heel pads had greater accommodation as tibia transmission that to head was reduced), which was in agreement the with passive increase in tibia acceleration and rate of loading. The viscoelastic component showed no significant correlations.

Comparing correlation of material properties with shock absorption parameters during walking and with passive properties as well as between passive properties and active tests would help to ascertain the materials contribution. Material properties that correlate simultaneously with passive and active impacts would make an important contribution to passive interaction. When passive and active impacts appear to be correlated with each other but not with materials passive interaction contribution should be considered. There are few material properties that can explain passive and active properties at the same time. Material properties determining peak force and rate of loading in both tests (Table 11.2) were hardness and time-to-peak displacement (Tdmax18). Initial stiffness and energy absorption at loading set 18 (100 kPa in 800 ms) correlated with tibia acceleration and upper transmission did with Kmax45. These material parameters should be considered for future experiments to differentiate between passive and active shock absorption. These parameters should be considered during footwear design

From these results, it could be thought that machine testing is able to predict walking and pendulum results. However, whilst this is true for general force and acceleration magnitudes (in agreement with Chapter 6 results) differences are found between head and tibia (Mc/Mt showed only a moderate correlation with Kfinal58) that suggest the presence of accommodation effects.

Table 11.2. Correlation of material properties with active and passive tests results

	ACTIVE	PASSIVE	
Fi	Kfinal18, Rebote28,	D300, D250, Kmax58, RT, tdmax41, Eabs24, tvisco45, tvisco48	FImpact
	hardnass, tdmax18		
Mc	Hardnass, kini18, Eabs18, Kmax45, kfinal58, Dmax48, tdmax43, Kfinal38 tvisco45	Tdmax18, Tdmax38, Tvisco 44	ap
Mt	Hardness, Kmax45, Kfinal58, Tvisco58 tvisco28	Dmax43, 44 and 48; Kmax18, Kmax41, tvisco48, RT, and tcodo18	at
	Kinl18, Eabs18		
Mc/mt	Kfinal58	R150 , kini18, tvisco45, tcodo42, tvisco48, Eabs18, kini44, 45, 48 and kini58, Kmax18, 41 and Kmax58	ap/at
	Kmax45		
ratac	Eabs18, Kfinal18, Rebote28,	Hardness, d300, d250, Kmax58, RT, tdmax41, Eabs24, tvisco45, tvisco48, tvisco58	Ratfim
	tdmax18		

11.4.3. Statement 3 - Conclusions

This statement argues that the effect of shoe on natural shock absorbing systems, such as walking kinematics and heel pad confinement, has greater influence on shock absorption in walking tests than underfoot materials and that mechine testing of materials fails to predict this effect. The proposed hypothesis states that if pendulum testing of different insoles in the same shoe identified differences in impact forces and accelerations then these differences could be attributed to the insole materials' role as a shock absorber or passive interaction. Differences detected only in walking tests were considered to be due to material's influence in accommodation. The results showed general differences between barefoot and shod in both passive and active conditions that indicate a significant shoe-effect. Barefoot was associated with greater and quicker impacts and accelerations, but lower impulse. However, greater energy absorption and stiffness but lower peak displacement was measured barefoot. By contrast, no significant differences in shift time were found between barefoot and different insoles. However, differences were also found between insoles supporting the importance of insole materials. All transmission parameters showed differences except wall - tibia transmission.

The properties of heel pad barefoot and shod were significantly different in the elastic behaviour showing an important effect due to materials, heel pad and interaction (mechanical coupling) whilst the viscoelastic behaviour was similar in both cases. The barefoot heel pad showed an initial elastic deformation at low stiffness that disappeared when shod. By contrast, the shoe-insole-heel pad system showed an elastic deformation component linked to final stiffness that was mainly dependent on shoe and insole properties. The elastic behaviour was mainly described by the maximal stiffness, which has a substantial influence of heel pad stiffness and passive interaction.

Impact mechanics was also different between barefoot and shod. Variance of shod impact and transmission parameters was explained by two principal components: One related to impact and wall acceleration parameters and the second to tibia acceleration and impact rate of loading. Impact forces barefoot were reduced by decreased stiffness and viscoelastic dissipation whilst peak time was governed by the initial deformation component at low stiffness. However, when shod impact forces and wall accelerations were reduced by increasing deformation at low stiffness which mainly depended in material properties but, in turn, tibia acceleration and the rate of loading increased. In this sense, decrease of the stiffness component reduced tibia acceleration and rate of loading. The elastic and viscoelastic behaviour were more human dependent and modified naturally by passive interaction depending on material parameters describing behaviour at initial loading. This result suggested that mechanical interface between heel pad and insole could be designed to compensate for increased tibia acceleration and rate of loading by deformation increase.

These results seemed to demonstrate an important contribution of materials to impact and acceleration at two levels: (i): determining stiffness and (ii) change in elasticity modifying the final compression part of the force-displacement curve to decrease peak forces and accelerations and, at the same time, increasing tibia acceleration and the rate of loading. Modifying the elastic behaviour (more participant dependent) by passive interaction at initial loading was supported by correlation with some material parameters describing initial behaviour (Kini18 and eabs18 correlated with both passive and active tibia acceleration). The viscoelastic component would be the result of combined heel pad and insole properties with little influence on the variance of impact mechanics.

Thus, materials determine the general magnitude of force and acceleration, whilst passive interaction determines accommodation that attenuates upper body transmission. In conclusion, material properties of hardness and stiffness obtained from machine testing are good predictors of the elastic deformation behaviour of the shoe-insole-heel pad system and of impact forces and wall accelerations. However, for tibia acceleration and rate of loading prediction human pendulum would be required. In general, at low and slow loads initial stiffness (Kini18, 28) of the insole material should not be greater than of the heel pad.

Contrary to what is proposed in Statement 3, these results appear to demonstrate that insole materials play an important role in shock absorption. However, to a great extent, passive interaction between insole and heel pad determines shock absorption. Passive interaction depends on the properties of the material and viscoelasticity (loading frequency sensitivity) of the heel pad and this could explain why machine testing is not able to predict material performance in use. This would require additional heel pad testing and further research for improved foot-insole interface design.

11.5. Statement 4. People accommodate to impact conditions. Proprioceptive model of Robbins and IBV modification.

As stated in the literature survey, accommodation to impact conditions is the most actively researched explanation for machine - human testing differences. According to Robbins (1989-1991), plantar discomfort due to pressure exceeding human tolerance provokes an avoiding behaviour known as accommodation and it is this that produces the differences. The human body has been described as a feedback controlled proprioceptive system in which different functions have been identified, namely:

- *Input sensing*
- *Control and objective contrasting*
- *Output to modify undesired input*

The study of these functions has raised many questions that are currently being investigated. An important part of these have been analysed in the present work. This statement required a complex analysis looking at the different aspects related to impact perception,

The human body has been described as a feedback system so input and output as well as system's goal were studied. In Robbins's model, input is affected by ground irregularities that modify the local plantar pressures and thus influence impact perception. In the literature, this is extended to perception of cushioning of impacts. However, the most recent evidence in this research, and early results reported in Chapter 6, suggest that proprioception depends on underfoot stiffness. At the same time, results obtained from the analysis of heel pad mechanics (Chapters 7 ands 10, and discussion with respect to Statement 3) indicate that underfoot stiffness comprises materials, heel pad and passive interaction. Passive interaction is the consequence of initial loading and mechanical coupling between materials and heel pad and their roles in accommodation was investigated in this work. As a result of this research, Robbins model (Robbins & Gouw, 1991) was modified and extended to consider the three components as determinants of impact input by two different mechanisms: change of impact mechanics and influence in perception by changes in underfoot stiffness. This was tested in this Chapter and the results are presented in three sections according to function within the model.

Nevertheless, general various aspects of Robbins' theory were first analysed. Initially, the question whether any accommodation took place was investigated and if it was the cause of the differed results between machine and human testing then several experiments were conducted to check the different components of the model. The role of plantar discomfort was then investigated, as this is a major component in Robbins model. Results from objective and subjective walking tests, pendulum and machine tests on ten different insoles were analysed.

11.5.1. Accommodation analysis

As already stated, ten insoles were tested into the same shoe in such a way that differences arising between insoles during walking should be due to insole material effect and could be due to the material alone, passive interaction or accommodation.

Chapter 6 described differences between material properties and head acceleration during walking, as well as between shod and barefoot behaviour, described as accommodation, in relation to underfoot stiffness. These results were obtained on different insert materials placed into the rear of the insole (using the IBV dynamic machine testing method). This did not allow passive interaction since there wasn't full contact between the heel and the material. However, in further work the whole insole was made of the same material so that the rear could adapted to the heel shape modifying behaviour under initial loading as a result. New material properties related to plantar pressure were identified using the new testing machine and are included in the study.

11.5.1.1. Material and Methods

Results of walking and passive tests were analysed to assess differences between insoles. Since the same shoe was tested any differences between insoles would be due to the insole alone be that by accommodation or materials shock absorption or passive interaction as discussed in relation to Statement 3. It was considered that if differences between insoles happen to occur only during walking tests and not in passive testing they were due to accommodation. At the same time, the relationship between impact forces and acceleration registered during walking with different insoles were studied as accommodation could be reflected as a reverse relationship between forces and accelerations (Chapter 6). Results of principal components and correlation analysis between shock absorption parameters were studied.

Finally, the comparison between passive and active testing was also done using ANOVA considering insole and method (active and passive) as factors to assess the ability of passive properties of underfoot materials in predicting results from walking tests. Any differences should be due to movement influence or/and accommodation. Assumptions for normality were tested for each variable using Lebene test for homogeneity of variance and Kolmogorov-Smirnov (K-S) test for normality.

11.5.1.2. Results, Discussion and partial Conclusions

As described in the previous section, differences found between insoles in walking tests implied that insole material had an important role. However, similar results were obtained from passive testing due to a substantial role of materials as shock absorbers or due to passive interaction but not for accommodation. Thus, differences in walking tests could be due to any of the three effects or any combination.

The principal component analysis implied that global variance of impacts during walking could be substantially explained by five components related to impact parameters and head acceleration, transmission and peak time, with different components for parameters normalised with respect to barefoot and absolute ones. A particularly interesting result was that one component referred to increase in impact parameters and acceleration but decrease in head - tibia transmission. Others reflected cases where greater tibia transmission yielded lower head - tibia transmission (nearer to barefoot which was the lower) which was consistent with previous results (Chapter 6) and interpreted as accommodation behaviour. Thus it was interpreted as accommodation was taking place.

Relationship of passive and active biomechanical variables

In a certain sense, pendulum impact testing can be considered as a machine test of participant-insole-shoe system without movement. This means that the role of participant and material properties and passive interaction could be measured simultaneously with no accommodation. In this test, properties related to shock absorption and attenuation were measured.

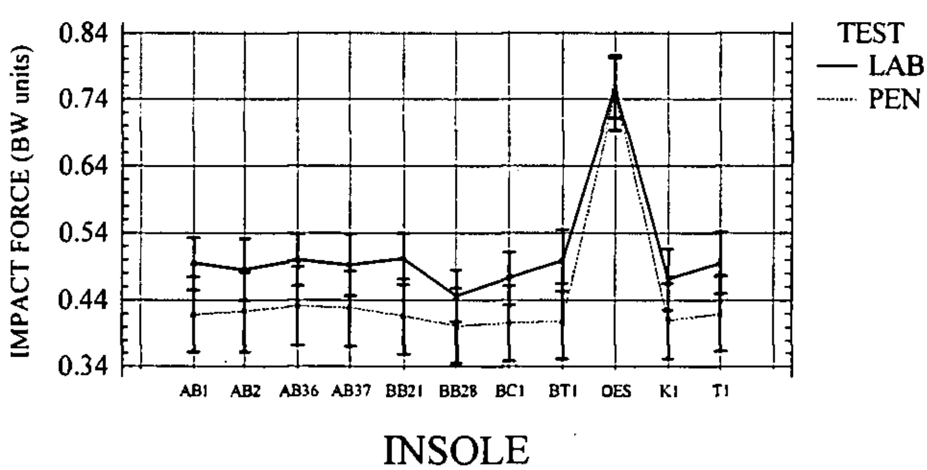


Figure 11.1. Impact force for both methods and insoles (Barefoot = des)..

The results of ANOVA analysis reflected significant differences in impact force between both methods for the insoles, but not for barefoot (Figure 11.1). Differences were also found for rate of loading (Figure 11.2) however, these differences were small except for barefoot (DES in the figures).

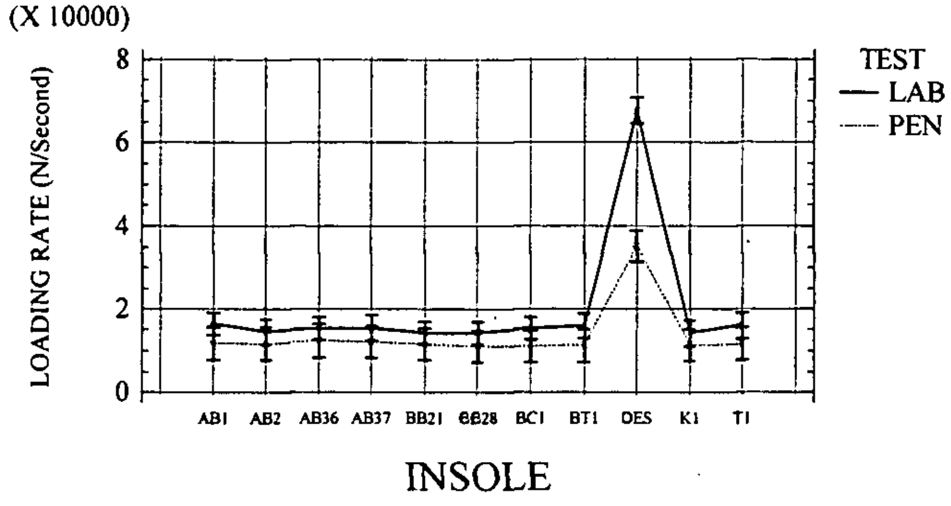


Figure 11.2. Rate of loading for both methods and insoles (Berafoot = des)

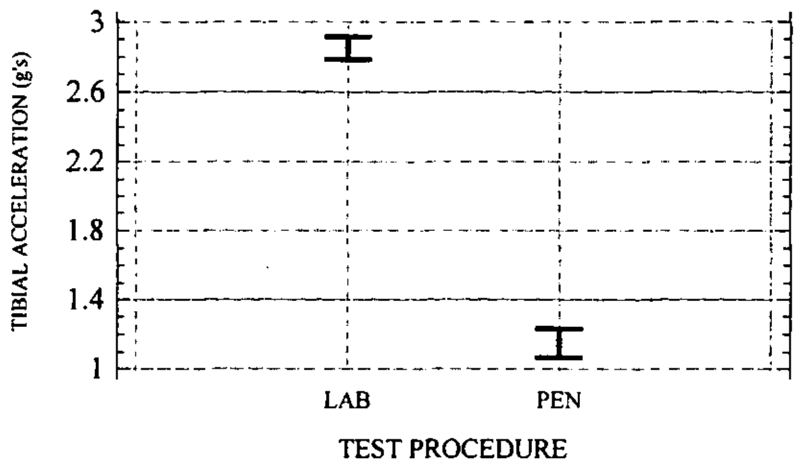


Figure 11.3. Tibial acceleration in both experiments

Nonetheless, the results of tibia acceleration (Figure 11.3) reflected greater significant differences. In general impact parameters and accelerations were greater in the laboratory (active test). However, results for normalised tibia acceleration (Figure 11.4) showed no significant differences whilst normalised peak force was greater in the pendulum test (Figure 11.5). However, tibia acceleration and normalised transmission were greater in the walking test (0.45 vs. 0.96).

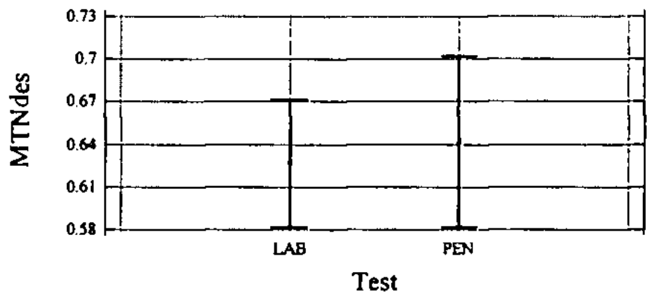


Figure 11.4. Normalised tibial acceleration (Mt/Mt barefoot) for both methods.

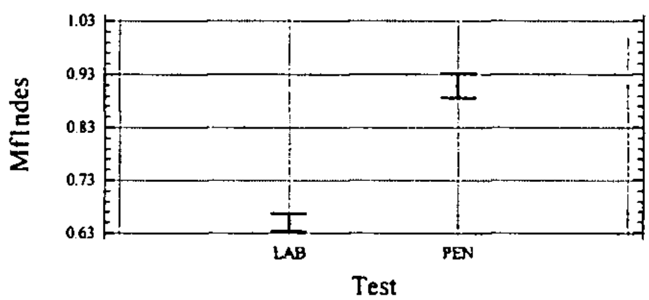


Figura 11.5. Normalised peak force (Mf1/Mf1 barefoot) in both methods

The comparison of walking and pendulum results reflected that for approximately similar peak forces and rate of loading. However, great differences in tibia acceleration and normalised values between insoles appeared in the active tests, which was attributed to acceleration due to leg oscillation during walking. In this sense, observed normalised peak force values were lower in the active than in the passive tests for insoles and this contrasts with results presented in Figure 11.1. Similar values were registered barefoot in both tests but lower for the insoles in the pendulum test which would mean greater ratio (i.e. lower reduction) in the active test contrary to that observed. In this sense, in Figure 11.5 participants variability is included, so kinematic accommodation could take place in response to impact. Greater tibia transmission in active conditions was also expected as greater peaks, and some influence of movement, in the acceleration of the leg as found (Figure 11.3). However, no differences in normalised tibia accelerations were found.

However, as described in section 11.4.2.2. results from correlation analysis between pendulum and walking tests showed that passive properties strongly predict walking results. It has to be remembered that only the insole material was changed so there was little need for the participants to accommodate. Perhaps for greater changes in impact force and rate of loading greater accommodation should be expected due to greater increase in tibia acceleration and greater decrease in upper transmission.

The results seem to support the notion that accommodation really takes place in the upper body showing an inverse relationship between accelerations registered in the upper and lower body. A relevant role of materials was also observed both as shock absorbers and related to passive interaction with heel pad.

11.5.2. Analysis of plantar discomfort role in humans accommodation

Robbins considers that when plantar pressures under the metatarsal heads surpass tolerance threshold, people initiate an avoiding behaviour by means of accommodation shifting pressure from forefoot to rearfoot. In this context, plantar discomfort and plantar pressures were analysed.

11.5.2.1. Material and methods

The results from subjective testing of discomfort in the body areas relative to foot plant were first studied. Results from pressure distribution analysis, principal component analysis and Pearson's correlation analysis with shock absorption and rearfoot movement were also analysed. The role of pressure distribution in comfort and impact perception was also studied by means of Pearson's correlation analysis using the mean rank from Kruskal-Wallis as an average measure.

11.5.2.2. Results, Discussion and partial Conclusions

Little evidence of avoiding behaviour was found from the results. There were few reports of plantar discomfort mainly located under the metatarsal heads. But no relationship either with insoles or participants was found. Whether it was because it did not occur or that participants avoided it by modifying walking kinematics was not clear. In the latter case, biomechanical consequences in the form of pressures, forces, accelerations, etc. should be found.

The analysis of pressure distribution reflected significant differences in the magnitude of global pressure between insoles. A general plantar pressure distribution was found showing that the pressure was always lower in the midfoot areas and greater in the central heel. In the forefoot the pressure was always greater under central metatarsal heads, and in some cases as high as in the lateral and medial heel. Nonetheless, some differences were found in the distribution between forefoot and heel pressures for some materials. The greater changes in pressure between materials occurred under the fifth metatarsal head and the rest of toes whilst in the rearfoot, great differences between materials were found under the central heel and, some, under the lateral (external). This would support Robbins theory if correlation analysis had reflected a relation between pressures under different areas indicating pressure shift. However, at all levels principal component analysis identified six components basically related to foot function and structure showing no pressures release mechanism (Chapter 9).

No significant correlations were found at participant level, which were expected to support avoiding behaviour, which depends on participants' pressure tolerance. Nor were any found at insole level. The relationship between different biomechanical variables was also studied.

The relationship between plantar pressure and impact parameters was studied in relation to, on one hand, to Robbins theory of plantar discomfort at accommodation onset and, on the other hand, through impact perception with heel pressures - as reported in the literature. Principal components were used for the analysis; the variables were at insole level.

All level (Critical Value, CV = 0.444)

Results (Table 11.3) showed low-moderate correlations

Table 11.3 Correlation between plantar pressures and impact parameters at all level

	Barefoot Impact	Impact	Transmission	Barefoot Transmission	Time
1sttoerelease	0.462	-0.490			
1Met head				-0.327	-0.282
Foot structure			-0.509	0.284	
Foot angle				0.368	
Heel					
Midfoot plant					

The noted moderate negativity (-0.509) between the component relating metatarsal heads, midfoot and internal heel in a foot structure component with shock transmission could support Robbins theory since this component reflects general distribution change in relation to the reverse relationship between head and tibia transmissions. No correlations with heel pressures were found

Insole level (CV = 0.889)

At insole level, both absolute and normalised head acceleration showed moderately positive correlation (> 0. 6) with external heel pressure (Table 11.4).

Table 11. 4. Correlation between plantar pressures and impact parameters at insole level

	MC	Mcnd	Mf1	Mf1nd	Mcmtnd	McMt	Mt	Mtnd	Ratac	Ratacnd
Mz1mt										
Mz5mt										
Mzmmt										
Mzplmc			0.709	0.641						
Mzplme										
Mzplmi										
Mzprdd										
Mzrdd										
Mztlc										
Mztle	0.668	0.648								
Mztli										

So an insole worn by an average human yields head accelerations related to external heel pressures probably due to impact force translation in heel pressures.

The correlation between pressure distribution and rearfoot movement parameters was sone to assesses the role of plantar pressure in walking kinematics as it is related to output it will be again considered later. Principal components were used for the analysis at each level whilst variables were used for insole level.

All level (CV = 0.444)

The results (Table 11.5) reflected low correlations except for foot position at initial contact with component showing transfer of pressures in the forefoot (r = 0.668) which seems to reflect a functional aspect, whilst plantar pressure distribution related to foot structure appeared correlated to knee angle at initial contact (r = 0.468).

At insole level (CV = 0.889)

Only internal heel pressure correlated with contact angle (0.639) probably related through impact force.

Table 11.5. Correlation between pressure distribution and rearfoot paremeters at all level.

	Ankle Inversion	Knee at ini	Max ankle flex.	Foot position
1mt			-0.452	
1s toe release				0.608
foot structure		0.468		
heel			-0.384	
midfoot				0.332
foot angle			0.356	

Correlation analysis with subjective testing and impact perception was also considered. Whilst humans related comfort to impact transmission to tibia agreed with Van Jaasveld et al. (1990) ***the analysis of comfort*** reflected that an insole worn by an average person showed moderately negative correlation of final comfort and internal heel pressure whilst initial comfort correlated with central heel pressure. In general, greater heel pressures were related to more comfortable materials that could be related to a safer and faster walking At participant level no significant correlations were found.

Concerning impact perception, at all levels the results reflected only low correlation (Table 11.6). They highlight the positive correlation between passive perception and heel pressures (0.333) and negative correlation with foot angle (-0.422). Thus softer passive impacts were perceived for greater heel pressures and lower foot angle.

Table 11.6. Correlation between Principal components of planter pressure distribution and impact perception.

	Passive perception	Active perception
1mt		0.334
1s toe release	0.357	
foot structure		
heel	0.333	
midfoot		
foot angle	-0.422	

But, **at insole level** (CV= 0.88) there was negative correlation between active perception and pressure under the rest of toes which correlated positive with passive results. This means that greater pressures were related to harder active impact perception. However, passive perception correlated positively at all levels with central heel and metatarsal heads and with rest of toes at insole level meaning that greater pressures were related to softer passive perception. These, to some extent, contradictory results are difficult to explain but, in general, harder impacts were perceived in active tests and have been demonstrated to be related to greater pressures under the forefoot (better sensors) and with impact parameters when participant - insole interaction was eliminated.

These results would strengthen the need for active testing of impact perception since in a passive condition there may be a masking effect that could result in misleading discomfort impact. It seems that active perception relies in impact events whilst passive perception participants make use of pressure sensors to perceive the strike. In the latter, there is no movement and no vibrations reaching the head (pendulum test is performed against a wall), thus participants make use of pressure sensors to perceive the strike.

In general, harder passive impact perception was related to heel plantar pressures decrease and forefoot increase which could support Robbins theory that human beings avoid forefoot discomfort by increasing rearfoot pressures when hard loads are perceived. This would be supported by results **at insole level,** greater forefoot pressures were related to harder active impact perception. However, passive perception correlated positively meaning that greater pressures were related to softer passive perception. Thus, participants seemed to sense impact by plantar pressures which could indicate that the translation of impact forces and accelerations (as showed by the correlation found between heel pressures and head acceleration) would be the determinant for perception and this stresses the need to understanding passive interaction that, by mechanical coupling, affects the contact surface and so plantar pressure distribution.

In conclusion, neither plantar discomfort nor pressure shift between forefoot and rearfoot were clearly observed but some correlations were found between shock absorption and impact perception with plantar pressures that could support Robbins theory which would require further work. In any case, plantar pressure distribution appeared to be important in the study of shock absorption. As a relevant result, because plantar pressures were thought best able to stimulate foot sensors they were used by participants to perceive forces, accelerations and movements. Thus foot and insole interaction should be deemed worthy of greater attention.

11.5.3. Goal of the system

According to Robbins model, the body has a control feedback system that compares actual input with a [safe] target input and signals the need to modify the input as necessary. The analysis of the system's goal was done considering the two most outstanding possibilities:

1. That goal of the system is to keep head acceleration, impact force and any other biomechanical variable at a safe value.
2. That accommodation is driven by comfort or a similar subjective perception about the shoe.

11.5.3.1. Material and methods

Results from ANOVA analysis of shock absorption were studied to evaluate the first possibility and the results of comfort testing and correlation analysis with different variables were studied to check the second possibility.

11.5.3.2. Results, Discussion and Partial Conclusions

The Variability of biomechanical properties was analysed in Chapter 9. Shock absorption results indicated small differences between insoles for impact forces and rate of loading (lower than 10% and COV around 25%). As expected, significant differences were found between participants for all the study variables.

On the other hand [as in Chapter 6] head acceleration showed significant differences between insoles. Thus the hypothesis that keeping head acceleration at a safe value would be the goal of the proprioceptive system was not confirmed and should be rejected. The acceleration oscillated between 1.92 g. and 1.24 g for participants, a large and unsafe range of about 35%.

However, the differences and range of values found for impact force and rate of loading were small as were the differences observed between participants as reported previously (Chapter 6). In this sense and – according to the literature and this research - impact perception is related to these variables and this result could imply that the system goal is to restrict rate of loading and peak force to a specific range, probably as a way of modifying impact perception to safe or comfortable values. This variable appeared also to be related to heel pressures. It could also be a mechanism to avoid undesired frequency content by controlling rate of loading since high frequency components of shock waves have been described to be more deleterious (Chapter 2).

On the other hand, the results reflected a low **role of comfort** in shock absorption. No significant differences were found in the comfort level between insoles. Participants showed, however, significant expected differences since every participant has his/her own comfort level. Nevertheless, a high range (from 1 to 5) of values was reported for comfort so comfort appears to be less a fine objective than the more global objective of walking.

In the correlation analysis, only a low correlation was found between final comfort and tibia transmission at participant level whilst at insole level there was a moderate negative correlation with internal heel pressure. Thus greater heel pressures were related to more comfortable materials.

This result was, to some extent, contradictory since greater heel pressures appeared simultaneously related to increased comfort (increase in tibia transmission) and harder impact perception (as described above). This could suggest that there are two different control mechanisms and two mechanical systems: one for comfort related to transmission regulation which is a long term result (one hour walking) and the other at short term for impact perception that regulates general level of forces and accelerations in a continuous feedback. This subject will be later discussed in detail.

The conclusion is that keeping impact force and rate of loading within a safe range in relation to either comfort or impact perception could be the goal of the system.

11.5.4. Sensing function. Analysis of impact perception

Impact perception is related to the sensing function of the proprioceptive system that registers and transmits the input to the control system. The various questions found in the literature concerning this aspect have been tested [Chapter 9] both the biomechanical and methodological aspects of human ability to perceive walking impacts as well as small changes in insole material. The influence of the reference condition in perception rating, the influence of experimental protocol and the comparison between perception collected during active and passive tests are major methodological aspects discussed in the literature. These questions are analysed below.

11.5.4.1. Materials and methods

The experimental work has been described in Chapter 9.

11.5.4.2. Results, Discussion and partial Conclusions

With respect to **Limitations of perception sensing**, results described in Chapter 9 presented a great range of impact perception scores by simple change of insole material with statistically significant differences between insoles. Barefoot impacts were perceived as harder, independently of the experimental method. These results made it clear that human sensing ability is not limited to the perception of walking impacts and small differences between impact conditions arising from insole material change. Differences were also found between participants as might be expected.

However, in passive testing no differences between insoles and a narrower perception range was found that could explain why results found in the literature, obtained using passive testing such as human pendulum (Lafortune et al., 1995a, b; Milani et al., 1997), limited perception sensibility only to running impact. Comparison between ascertained perception in active and passive conditions showed that for very low differences in impact forces and rate of loading (Chapters 9, 10 and 11.4) perception appeared to be harder in active tests than in passive tests with no significant correlation between them. This is important since work found usually referred only to passive methods thus perception could be minimised. In this sense, harder perception was observed in the active testing when participants were not informed of the reference condition and used the shoes for a longer time than in walking trials when shoes were worn for a few minutes only and when participants were aware that the reference condition was being re-tested. Thus, active testing would be recommended for impact perception analysis.

At the same time, the experimental methodology was showed to have great influence in final results. It seemed that in the active experiment, perception was more participant dependent (significant differences were found) due, either to participants having more information to judge impact perception based on their past experience (more trials and the walking trial experiment had been already carried out), or, to participants being more relaxed regarding rate perception since they were not informed about reference condition.

Important results about the use of reference condition were observed. The first shoe tested in active conditions was always reported as softer. Thus, methods in the literature using reference scaling with respect to the first condition could introduce an important error. However, this result was not confirmed in passive testing which is the method most commonly found in the literature. On the other hand, reference insoles were shown not to affect perception in the following tests.

In conclusion, future experiments should adopt the following recommendations (see Chapter 4):

- Use active testing methods
- Simplify the test to a single task
- Not use the first test condition as reference nor for or scaling
- Retest reference condition every few tests

11.5.5. Input

The interface between walker and ground determines the input for the proprioceptive system and it is this input that defines the impact perception and comfort. This interface has three different components: wearer, shoe and insole materials and their interaction. The interface defines a series of biomechanical variables, such as forces and accelerations that enter to the system, and describe human walking at control level. However, it is not clear which biomechanical variables or combination of variables comprise the system input and how they depend on underfoot properties. The same applies for material and heel pad properties: It is not clear, which material or heel pad properties or combination of properties relate to impact perception. This is very important since knowing which variables govern impact perception would allow perception to be controlled and the interface (shoes, surface, etc.) designed to modify perception as desired.

11.5.5.1. Material and methods

The relationship between impact perception and biomechanical variables related to shock absorption, rearfoot movement and pressure distribution were first studied to establish input variables. Pearson's correlation analysis was done between impact perception and principal components obtained from walking tests and with variables at insole level. Further correlation analysis was done with material and passive properties to study the role of underfoot materials, heel pad and passive interaction.

11.5.5.2. Results, Discusion and partial Conclusions

The results are presented in two parts:

1. Biomechanical variables
2. Interface variables: human, shod heel pad and material variables.

In relation to the ***relationship between Impact perception and biomechanical variables***, the results of correlation analysis between impact perception in passive and active conditions with principal components identified from shock absorption, rearfoot movement and pressure distribution are presented (Table 11.7).

Teble 11.7. Correlation of perception with biomechanical veriables

	Pessive perception	Active perception
Impact		-0.395
Barefoot Impacts	-0.444	-0.594
Transmission	-0.225	
Barefoot transmission	-0.326	-0.240
Time		
1 met. Head		0.339
1st toe release	0.357	
Heel	0.33	
Foot angle	-0.422	
Foot structure		
Midfoot		
Ankle inversion		
Knee at initial	-0.232	
Max ankle. flex	-0.208	
Foot position	-0.424	

At all levels, active perception was more frequently correlated with impact than with pressure parameters. They were negative, as expected, so the greater the impacts the harder they were perceived which strongly support the reliability of the method for impact perception testing used in this research. The greater correlation was with impact parameters normalised with respect to barefoot which was also the greater for passive perception. The nearer to barefoot the impacts the harder they were perceived. Low negative correlations were found with transmission.

Passive perception showed some low correlations with pressure principal components that were positive with heel pressures and first toe transfer component and negative with foot angle pressures component. This could show that, in the absence of active stimuli, pressure (force/surface) was used for sensing.

Considering rearfoot movement, low negative correlations with passive perception were found to be higher negative with foot position at initial contact in such a way that the flatter the foot at initial heel strike the softer the perceived peak. This seems rather logical but it is not clear whether foot position was a response to regulate impact perception or the cause of soft perception.

At insole level, no significant correlations with plantar pressure were found. However, there were many correlations of impact parameters with active perception. Correlation was negative with rate of loading and tibia acceleration but positive with head/tibia transmission thus for an average participant harder impacts were perceived when quicker impacts occurred and less tibia acceleration reached the head in accordance with literature reports (Hennig et al., 1996; Lafortune et al., 1995b; Milani et al., 1997).

Considering the ***analysis of correlation of pendulum results with perception parameters***, *at all level (CV = 0.444)*, (Table 11.8) the impact-wall component (a high score of impact force) showed moderate negative correlation with passive and active perception, the correlation being greater with active perception. The rate of loading and tibia component, however, showed greater correlation with passive perception, which would be consistent with literature where reported studies, in general, were conducted under passive conditions showing greater correlation with rate of loading.

Heel pad principal components showed moderate negative correlation (r = -0.6) of active perception and of elastic deformation which was low with passive perception. The initial-maximal stiffness component correlated only low negative with passive perception whilst the viscoelastic component showed no significant correlations. These results seem to support the existence of a control system for impact regulation linked to the elastic deformation and impact parameter components in such a way that greater shod heel pad elastic deformation component (lower peak deformation) is related to harder active impact perception.

Table 11.8. Correlation results between impact perception and passive shod heel pad results

	Passive	Active	Active long term
Impact-wall comp.	-0.437	-0.613	-0.620
Rate - tibia comp.	-0.451	-0.249	-0.271
Elastic deformation comp.	-0.394	-0.636	-0.622
Initial-maximal comp.	-0.418		
Viscoelastic comp.			

In the barefoot analysis (CV = 0.889). The results (Table 11.9) showed moderate positive correlation of passive perception with peak force (0.75) and wall acceleration (0.686). Time-to-peak correlated moderate negative with active perception in long term trial (-0.70), but no correlations with heel pad principal components were observed. This confirmed that human-characteristics have little influence in impact perception. Human-characteristics influence only impact parameters - both force and rate of loading resulting from passive testing.

Table 11.9. Correletion results for berefoot heel ped

	Passiva	Active long term	Active
Fimpact	.750		
RATFIM			
TFimpact		- 0.70	
Apmax	.686		
ATMAX			
facelas			
facni			
facvisco			

At insole level (CV = 0.889). The results (Table 11.10) showed only a few significant correlations, similar at all levels; only from walking trial experiments were perception correlations negative. Moderate negative correlation of active perception with peak force (-0.669), wall acceleration (-0.637) and elastic component (-0.690) were found when participants used the insoles.

Table 11.10. Correlation results at insole level

	Passive	Active Long term	Active
Fimpact		-.669	
TFimpact			
RATFIM			
APMAX		-.637	
ATMAX			
FACELA		-.690	
FACINI			
FACVIS			

The results of the ***analysis of the role of material properties*** showed that active perception was highly correlated (> 60% of variance explained) with many material parameters obtained from dynamic impact test, static compression and dynamic cushioning. It was positive with d250, d300, tan5, Dmax18, Eabs18, tdmax41 and negative with r150, hardness, Kfinal18, Kini18, Kini28, RT, tdmax38, tdmax43, tdmax44, tdmax48. In walking trials it correlated positive with dmax28 and negative with rebote28, which are probably related parameters and describe long term properties. However, passive perception showed only moderate positive correlation with tcodo43 and negative with tcodo45 and tvisco38 demonstrating a low role of material properties.

Different parameters and type of parameter correlated with passive and active perception. Active perception was correlated with elastic global and initial loading and viscoelastic parameters in a logical way: harder, stiffer global at initial loading and less energy absorbing materials yielded harder impact perception in active conditions whilst in long term the recovery-ability of materials was also effective in predicting impact perception. In passive conditions few parameters were correlated to perception. Of these, hardness showed correlation with peak force and rate of loading both from active and passive tests and with Kini18 and Eabs18 with tibia acceleration.

11.5.5.3 Discussion and Partial Conclusions

Results from Chapters 6 and 9 suggest that accommodation is related to underfoot stiffness as a result of the combined effects of insole material, human characteristics and passive interaction. In this sense, the results of principal components and correlation analysis described above suggested the existence of two different mechanical systems acting in different ways in impact perception and, therefore, in shock absorption.

Considering biomechanical variables from *active tests* at all levels. When interaction between participant and insole was considered, active perception appeared to be explained by impact parameters in a logical way: the greater the impact force and acceleration the harder the impact was perceived which was related to flatter foot at initial contact and considered a mechanism of regulation of shock absorbing behaviour. The greatest correlation was with impact parameters normalised with respect to barefoot which was also greatest for passive perception. The nearer to barefoot, the harder they were perceived. By contrast, passive perception appeared more related to plantar pressures, being high for harder perceptions. At insole level when human variability was eliminated, quicker force peaks and higher acceleration yielded harder active perception but positive correlation with upper transmission, which decreased. This behaviour is the accommodation.

Passive properties were shown to play an important role in impact perception. Greater passive impact forces, rate and accelerations as well as lower deformation yielded greater perception of impact both passive and active. Impact-wall component which is positively correlated with the elastic deformation component showed greater correlation with active perception whereas passive perception depended more on rate of loading of tibia and initial-maximal components which are correlated with each other, this result is in agreement with literature reports that correlate perception with rate of loading (Lafortune et al., 1995b; Milani et al., 1997).

In general, these results with active tests confirm literature findings that show that parameters related to frequency content of impacts are better for predicting impact perception (Hennig et al., 1996; Lafortune et al. 1995b; Lake & Lafortune, 1997).

The elastic deformation component showed moderate negative (-0.636) correlation with active perception and low correlation with passive perception. The initial-maximal stiffness component of the shod heel pad, which is related to transmission component in active tests, showed only low negative (-0.418) correlation with passive perception in contrast to active testing that showed a negative correlation between upper body transmission and active impact perception. This component has been previously described as being related to passive interaction which influences pressures beneath the heel by either confinement or contact area increase. Thus, a stiffer shod heel pad could also result in greater plantar pressures that, in passive testing, could lead to perceived harder impact. This requires further research.

Analysis of results barefoot (human characteristics) showed that passive peak force and acceleration were related positively with passive perception so that people bear greater impacts barefoot (stiffer heel pad as seen in Chapter 7) perceived softer impacts. This result could mean that participants with a stiffer heel pad experienced lower heel pressures when shod in passive tests due increased confinement rather than contact area increase as result of mechanical coupling with the insole and hence softer impact perception. It was established that heel pad stiffness has significant influence in the initial-maximal stiffness component of the shod which, in turn, appeared to be correlated with passive perception. As stated above, a stiffer shod heel pad could result in greater plantar pressures leading to a perceived harder impact. However, those participants with a stiffer heel pad appeared to experience lower heel pressures when shod and softer impact perception. These results seem contradictory as barefoot heel pad stiffness was linked to shod initial-maximal stiffness in a positive way. This requires further research.

But at insole level, elastic behaviour and wall acceleration showed negative correlation with active perception, which could reflect the role of elastic properties of materials in impact. This is supported by correlation analysis with *material properties* that showed material parameters such as elastic global and initial loading and viscoelastic account for more than 60% of variance of active perception. Logically, harder, stiffer global at initial loading and less energy absorbing materials yielded harder impact perception in active conditions whilst the long term recovery-ability of materials also influenced extent of residual displacement and greater this is, the harder the perception of impact. However, only a few material parameters were correlated to perception in passive conditions. Only hardness, which showed correlation with peak force and rate of loading both from active and passive tests as well as kini18 and eabs18 with tibia acceleration, correlated with perception data from active and passive tests.

These results would confirm an important role of materials and participant properties and the role of passive interaction in impact perception in determining impact and transmission parameters that seem to govern perception. In accordance with traditional theories (Hennig et al, 1996) hardness is a very important parameter for impact perception.

The elastic deformation component of the shod heel pad which is strongly influenced by hardness and elastic and viscoelastic materials properties was related to passive impact-wall parameters and, further, determined active impact parameters and head acceleration results. As a result, it showed negative correlation with impact perception in such a way that the stiffer, less absorbers and harder underfoot materials resulted in greater elastic deformation component leading to greater impact forces and accelerations both in passive and active testing as well as harder impact perception.

On the contrary, the initial-maximal component which depended on barefoot heel pad properties, material properties at initial loading and passive interaction appeared to be correlated to passive tibia acceleration and rate of loading and active shock transmission through participants characteristics. It correlated with passive perception being harder than the impact perceived for stiffer shod heel pad. At the same time, passive impact perception correlated negatively with heel plantar pressures, which could be related to the influence of passive interaction - initial-maximal passive component - in heel pressures.

From the above results a general qualitative model comprising two mechanical systems each acting in a different way could be proposed for impact perception and shock absorption. This is described in next sections.

11.5.6. Output

The output of the system would include those actions undertaken to correct undesirable inputs according to the target established by the control system. From impact perception results, a relationship between perception and foot position at initial contact was found which could represent a kinematic accommodation to impact severity, agreeing with papers found in the literature search (Clarke et al., 1983b; McMahon et al, 1987; Robbins et al., 1988).

11.5.6.1. Material and methods

Rearfoot variables at initial contact were analysed by principal component methods. ANOVA results were examined together with Pearson correlation analysis with impact perception and shock absorption variables. At insole level, Pearson correlation was done between variables.

11.5.6.2. Results, Discussion and partial Conclusions

The ANOVA results (Chapter 9) reflect that rearfoot movement depended mainly on the participant, but some significant differences were found between insoles which could be due to accommodation. Principal components identified patterns of movement related to walking kinematics. Patterns of movement extracted from the analysis showed foot position and knee flexion at initial contact as well as maximum ankle flexion and inversion movements.

Correlation analysis between rearfoot and impact parameters was used to assess possible kinematic accommodation to impact conditions.

At all level (CV = 0.44).

Table 11.11. Correlation between impact and rearfoot parameters at all level

	Ankle Inversion	Knee at ini	Max. Ankle flex.	Foot position
Impact				0.618
Impact barefoot	0.260			-.334
Transmission	-0.500	-0.433		
Transmission barefoot		0.588	0.429	-0.304
Time				

Impact parameters (Table 11.11) were correlated moderate positive with foot position at initial contact showing that a flatter foot was associated with lower and slower impacts as accommodation mechanism reducing impact forces (consistent with the literature). At the same time, moderate negative correlation of ankle inversion and transmission component was observed reflecting that greater ankle inversion component was associated with greater head transmission and lower tibia transmission (tibia transmission showed a positive contribution to this component whilst head transmission contributed negatively, see Chapter 9). These results suggest that there exist two separate regulatory mechanisms: one for transmission and other for impact forces. Knee position at initial contact and maximal ankle flexion also appeared to be related to transmission components.

Insole level

Only negative correlation of minimum knee flexion with absolute and normalised rate of loading (-0.736 and -0.755 respectively) and with normalised head transmission (-0.776) were found. As expected, this showed that if participant variability was eliminated, quicker impacts are associated with a reduced bent knee at initial contact.

These results suggested that kinematics adjustments really occurred to regulate impacts and transmission. Foot position and knee flexion were used to control impact parameters and accelerations whilst ankle inversion movement was related to upper transmission regulation. Thus lower and slower impacts and accelerations were achieved by flatter foot and extended knee bend at initial contact, whilst transmission between tibia and head was reduced by reducing ankle inversion. The latter result, despite of being relevant has not been reported by other workers.

11.5.7. General discussion and conclusions

The analysis of the results presented so far in relation to Statement 4 described relevant conclusions in the research fields of human accommodation and impact perception. Differences between insoles were found in forces and accelerations both in passive and active testing. This, in general, linked Statements 3 and 4 since it was not easy to discem the role of materials: was it shock absorption, or its effect on accommodation or a combination of both?.

The results seem to support the notion that **accommodation** in the upper body really takes place: principal component analysis of active impact parameters showed reverse relationship between tibia and head transmission. Besides, ANOVA results reflected, as expected, that barefoot yielded greater impact forces and accelerations as well as greater rate of loading. However, as described in Chapter 6, forehead transmission was lower barefoot, probably as a result of accommodation. Materials' contribution confirmed a reverse relationship between underfoot stiffness and head transmission. Underfoot stiffness is a function of materials' properties, heel pad characteristics and their passive interaction.

Concerning **impact perception**, the results of this research demonstrate that human sensing ability is not limited to perceive walking impacts or to small differences between impact conditions arising from a change of insole material. Barefoot impacts were perceived harder which (irrespective of the experimental procedure) appeared to have a great influence in results. In passive testing no differences between insoles and a narrower perception range were found and these could explain why other workers (Lafortune et al., 1995a, b; Milani et al. 1997) found perception sensitivity limited only to running impact because they mainly used passive human pendulum testing.

At the same time, comparison between ascertained perception in active and passive conditions showed that for very low differences in impact forces and rate of loading perception (Chapters 9 and 10) appeared to be harder in active tests than in passive tests with no significant correlation between them. The first insole tested in active conditions was always reported as softer and, therefore, should not be used as reference for scaling. However, this result was not confirmed in passive testing which is the method most commonly found in the literature. Active tests, therefore, were demonstrably better for assessing impact perception.

Little evidence of **avoiding behaviour** as described by Robbins was observed. Plantar discomfort appeared to occur in a few cases. Besides, the results of principal component analysis and correlation analysis of plantar pressures with impact parameters and rearfoot movement at participant level reflected only functional, structural or geometrical relationship between plantar pressures. A moderate negative correlation between the principal component reflecting general distribution in the function of foot structure and the component showing reverse relationship between head and tibia transmission was found. This could support Robbins theory linking plantar pressures and shock absorption. It could indicate, however, that there is a relationship between the patterns of foot structure (plantar pressure distribution) and accommodation: more supinated feet seemed to show greater tibia transmission but lower upper transmission.

The analysis of the different functions of the Robbins proprioceptive together with shod pendulum results (Chapter 10) and Statement Three conclusions (Section 11.4) were the basis for a new model. This model includes two mechanisms each acting in a different way for impact perception and shock absorption. These mechanisms differ in the goal, input variables and regulation system as described below.

Mechanism 1

This mechanism depends on the elastic deformation of the low stiffness component of the shod heel pad which, in turn, is mainly governed by the hardness and elastic properties of the insole materials and to a lesser extent by the final stiffness and viscoelastic properties of these materials. This low stiffness component defines the general magnitude of impact forces and accelerations both active and passive with increase in upper body transmission as accommodation (moderate negative component of tibia - head transmission for regulation). The resulting forces and accelerations show a negative relationship with active impact perception. This was described as a short-term goal that ensured that harder and stiffer underfoot materials resulted in a stiffer mechanism leading to greater impact forces and accelerations. But decreased upper body transmission was identified as accommodation. As a result, it led to harder impact perception during walking as a short-term feeling. Finally, a flatter foot and increased knee bend at initial contact are found to regulate impacts to soften the impact perceived.

Since differences were found between insoles, keeping head acceleration and impact forces at a safe value was not the goal of the system. However, small differences were found for the impact forces and rate of loading that have also been related to impact perception and heel pressures and that controlling the rate of loading to keep impact perception at safe and comfortable levels could be the goal of the system

Active perception appeared to be related to impact and transmission parameters whereas passive perception was related to plantar pressure. ***The role of pressure in impact perception*** was low as reflected by correlation results. In general, harder passive impact perception was related to heel plantar pressures decrease and forefoot increase which could support Robbins theory that people avoid forefoot discomfort by increasing rearfoot pressures when hard loads are perceived. But **at insole level,** however, greater forefoot pressures were related to harder active impact perception [again supporting Robbins theory] whilst positive passive perception correlated meant that greater pressures were related to softer passive perception. In general, human beings seemed to sense impact by plantar pressures, which could indicate that the translation of impact forces and accelerations would be the determinant for perception.

Consequently, the considered aim of this mechanism was to minimise the perceived active impact by reducing impact forces through a flatter foot and increased knee bend at initial contact to engender comfortable walking conditions. Thus it determines the general mechanical behaviour of humans to impact perception as a short-term goal, regulated by kinematic adaptation.

Mechanism 2

This mechanism depends on the initial-maximal stiffness component of the shoe-insole-heel pad combination, which was less dependent on the properties of insole materials. It was mainly determined by barefoot heel pad stiffness, insole properties at initial loading and the effect of passive interaction with insole that alters the initial loading behaviour of the global system. Heel pad properties and hence the mechanism depends on personal characteristics of gender, age and obesity. Greater initial-maximal component (stiffer heel pad) yielded quicker impact and greater tibia acceleration in passive conditions resulting in greater tibia transmission and lower head transmission during walking. This component was related to the regulation of upper body shock transmission by means of passive rate of loading and tibia acceleration and it determined the lowering tibia-head transmission for tibia transmission increase during walking. Ankle inversion was observed as the regulator for upper body transmission.

This mechanism appeared to be related to long term perception of footwear. It correlated with passive perception with harder impact perceived for stiffer shod heel pad probably due to its influence in heel plantar pressures. ***The analysis of comfort*** showed that participants related comfort to impact transmission to tibia whereas, in general, insoles generated greater heel pressures with the materials rated as more comfortable by wearers. The comfort effect was in shock absorption was small and observed only with tibia transmission whilst greater heel pressures were related to more comfortable materials.

Passive interaction is the mechanical coupling between heel pad and insole and is a function of the different viscoelastic nature. Therefore, the stiffness of both change independently according to the applied rate of loading. Thus, either heel pad confinement or compression against the insole (altering the contact area) occurred and the plantar pressures that seems to govern the passive perception, that appeared to be correlated negatively with foot position at initial contact, was suggested as a long term goal for comfort or health.

Thus, to reduce impact severity perception the human body behaves as a mechanism with greater forces and accelerations associated with stiffer and less energy absorbing materials and regulated by knee movement and foot position However, the upper body reacts to minimise shock transmission from tibia by ankle inversion to augment long term comfort. The human body has a feedback control system that responds in a way that sometimes affect transmission regulations to the mechanical system (principal components include impacts and reverse transmission) giving rise to the differences found between machine and human testing. These are two independent mechanisms. Thus, if the lowering of upper transmission was sufficiently large this mechanism could influence head acceleration and mask the effect of the first mechanism and introduce differences between machine and human testing.

It follows, therefore, a **conclusion** that proposes two independent mechanisms each with separate goals and regulatory mechanisms differences between machine and walking tests is not unexpected. Machine tests may predict results with respect to the first system but whenever the second is activated differences will appear and hence passive human testing would be required.

Thus, soft materials will reduce impact forces and accelerations by increased deformation and energy dissipation but they also increase head transmission due to accommodation (System 1) This increase could be compensated by System 2 if heel pad properties are known and mechanical coupling with insole designed to increase the initial-maximal component. Different stiffness systems could also be excited at different frequencies with the underfoot system acting as a filter. Given that the effect of impact loading depends on frequency, the analysis of this would be very important. If it really occurred, the frequency content of signals in each system would be very important and should be investigated.

From these results, high-energy absorption materials would be preferred; such materials should be capable of increasing elastic deformation without significant increase in accompanying initial-maximal stiffness. Materials systems to be developed as heel contact materials should retain a low stiffness initial response by means of different stiffness layers or be suitable for heel cup design. Those materials should present an initial stiffness similar to that of the heel pad. Based on these suggestions a new generation of foot friendly materials could be devised.

11.6. CONCLUSIONS

The aim of this research was to investigate the different Statements that have been proposed to explain the differences found between the properties of shock absorbing systems measured in machine tests and those obtained from walking tests. Conclusions arise from work reported in this thesis are presented in so far as they refer to each Statement. The results clarify many of the current research issues associated with shock absorption and represent a significant advance in this important field of knowledge. Nevertheless, in investigating the four Statements several more general shock absorption issues were identified and discussed and recommended for further research. These are presented as general conclusions.

Nonetheless, the ultimate purpose of this work was to accelerate the research in shock absorption. This research is currently in abeyance due to a range of conflicting concepts held by the research community. The goals of this research have been achieved along with answers to many related issues and questions. This research has identified further work need to progress this area of biomechanics and a list of important recommendations for further research is included as part of these conclusions.

11.6.1. On statement 1. There is no accurate simulation of impact loading in machine test

The results of this research have demonstrated that accurate simulation of loading history (as with the IBV method for impact testing of materials) improves the ability of material testing to predict the walking performance of heel insert materials in terms of force and acceleration that are valid up to the knee, but not for whole insoles. Between the knee and the head accommodation occurs and this introduces differences depending in underfoot stiffness. This work has been published in the Journal of Foot and Ankle International (Forner et al, 1995).

Thus, for heel inserts, in general, materials with a high loss tangent and relatively low rigidity should be preferred for reducing impacts transmissions from ground to the tibia and forehead. Optimum rigidity would yield diminished transmission to tibia without lowering impact proprioception and, therefore, without increasing the transmission of impact from tibia to forehead.

11.6.2. On statement 2. Materials degrade in use with loss of some of the properties that are measured by machine testing

As part of this research, a new testing method was developed to measure the ability of materials to recover after compressive loading by simulating plantar pressures. The procedure included two tests: (i) rapidly applied high pressure (simulating pressures under the heel and first toe) and (ii) slowly applied low pressure (simulates pressures under the 5th metatarsal heads). IBV offers this procedure commercially as a service to the footwear industry.

Residual compression appeared related to long term performance of materials in a way that greater residual displacement was related to perception of harder impacts as well as of harder under-heel material. Besides, residual displacement increase and resulting material thickness decrease but without collapse give rise to rate of loading increase as well as harder impact and harder material perception since any correlation with either impact forces or accelerations was found. A paper in this subject is currently under preparation

It has been shown that loss of properties with use influence shock absorption by modifying impact perception and that frequency of impacts influence accommodation. However, support for this being the source of differences between machine and walking test results was not clearly established. Nonetheless, the need for long term material testing to improve the predictability of materials' performance in use was clearly established and further research is recommended. In general, research in shock absorption studies should also include long-term properties of materials.

11.6.3. On statement 3. Shoe influence in walking kinematics and heel pad confinement has greater effect on shock absorption than the properties of insole materials.

Contrary to what is proposed in this Statement, the results of this research appear to demonstrate an important role for insole materials in shock absorption. However, it was not clearly established whether it was as shock absorber. A complex mechanical system was proposed in which the materials' influence as shock absorbers and in passive interaction with heel pad, both of which influence accommodation.

In this thesis passive interaction has been defined as the mechanical coupling between underfoot materials and heel pad that determines initial elastic deformation and subsequent increased stiffness under loading. This could originate either through heel pad confinement into the insole or by compression against the insole, increasing contact area and thus reducing pressures. Passive interaction depends on the relative properties of the material and the heel pad, which could explain why machine testing is not able to predict the performance of a material in use. This means that further heel pad testing is needed.

Generally, the properties of the materials determine the magnitude of force and acceleration, whilst passive interaction determines accommodation that then attenuates upper body transmission. In conclusion, the material properties of hardness and stiffness obtained from machine testing offer a good prediction of the elastic deformation behaviour of the shoe-insole-heel pad system and of impact forces and wall accelerations that result. However, to predict tibia acceleration and rate of loading human pendulum testing would be required. In general, initial stiffness of insole materials at slowly applied low loads (Kini18, 28) should not be greater than that of the heel pad.

In conclusion, materials have a relevant role in shock absorption and future work should consider frequency behaviour and the human/material properties relationship for a better understanding of shock absorption.

11.6.4. On statement 4. Humans accommodate to walking conditions according to a proprioceptive system model.

Conclusions thus far, connect Statements 2, 3 and 4 because, on one hand, the influence of long term loss of materials' properties appeared to be related to impact and materials perception and, on the other hand, the role of materials in shock absorption ultimately affects accommodation.

In connection with Statement 4 several issues were investigated and the conclusions with reference to the general issues are presented first and then the proprioceptive model.

Accommodation was observed to occur in the different experiments. With respect to the role of plantar pressure in accommodation (after Robbins) no plantar discomfort was reported and little evidence of avoiding behaviour detected. In general, rearfoot and leg movement were a measure of patterns of movements that could be determined by impact parameters and sensed through the resulting plantar pressures that appear to be better stimuli than forces and accelerations for human sensors and, therefore, people would rely on these for perception. In this sense, a general plantar pressure distribution pattern was described. The insole material modified the existing global pressure under the fifth metatarsal heads, central and external heel.

With respect to impact perception, the results of this research demonstrate that human sensing ability is not limited to perceived walking impacts or to small differences between impact conditions arising from a change of insole material. Active tests were demonstrated to be better for ascertaining impact perception and the future active testing should incorporate the following recommendations:

- Simplify the test to a single task
- Do not use the first test condition as reference neither for scaling
- Retest reference condition every few tests

Accommodation was confirmed as the source of differences between machine and walking tests of shock absorbing materials. The human body has a feedback control system formed by two independent mechanisms that respond in a way that sometimes affect transmission regulations to the mechanical system giving rise to these differences.

One mechanism depends on the elastic deformation at low stiffness component of the shod heel pad that is governed mainly by hardness and elastic properties of materials and to a lesser extent by the final stiffness and viscoelastic properties of materials. This defines the general magnitude of impact forces and accelerations with increase in upper body transmission as accommodation. The considered aim of this system was to minimise the perceived active impact by reducing impact forces through a flatter foot and increased knee bend at initial contact a more pleasant walking. Thus, this system determines the general mechanical reaction of human beings to impact perception regulated by kinematic adaptation as a short-term goal.

The other mechanism depends on the initial-maximal stiffness component of the shoe-insole-heel pad system that was less dependent on materials properties. It was mainly determined by barefoot heel pad stiffness, insole properties at initial loading and the result of passive interaction with insole that alters the initial loading behaviour of the global system. Heel pad properties and, hence the mechanism depends on individual characteristics as gender, age and obesity. It was related to the regulation of upper body shock transmission by means of passive rate of loading and tibia acceleration. Ankle inversion was observed as the regulation for upper body transmission. This mechanism appeared to be related to long term perception of footwear as ***comfort*** was related to impact transmission to tibia.

In conclusion, if a system has two independent mechanisms each with independent goals and regulatory protocol, then different results from machine and walking tests are likely. Machine tests may predict results with respect to the first mechanism but when the second mechanism is activated differences will become apparent and for this reason, passive human testing is necessary.

It follows, therefore, that high-energy absorption materials would be preferred. This type of material should be capable of increasing the elastic deformation that reduces impact forces and accelerations and energy dissipation but without significant increase in accompanying initial-maximal stiffness thereby minimising head transmission due to accommodation. Any tendency to increase head transmission could be compensated by the other system if heel pad properties are known and mechanical coupling with insole designed to increase the initial-maximal component. This offers opportunities for the development of a new generation of underfoot materials.

11.6.5. General Conclusions

To explore the different research issues, basic knowledge was reviewed and new concepts studied with significant results. These results are presented as general conclusion with reference to:

- General knowledge about walking biomechanics
- Heel pad mechanics, barefoot and shod, and dependence on human characteristics.
- Passive Interaction
- Material testing

11.6.5.1. Walking biomechanics

The analysis of different biomechanical aspects of walking advanced the knowledge of human gait. Interesting conclusions about plantar pressure distribution, rearfoot movement and shock absorption were obtained.

ANOVA results reflected greater differences between humans than between insoles, which is normal since individual differences due to walking pattern, speed of walking and other aspects are likely to be greater than differences due to insole material. On the other hand, as expected, barefoot tests yielded greater impact forces and accelerations as well as greater rates of loading.

Considering plantar pressure distribution. Insole materials modified global plantar pressure magnitude. There was a general pressure distribution pattern as already reported (Chapter 8) and as found in the literature. The higher pressures were found under the heel, especially in the central heel (TLC), and in the central metatarsal heads (MTM), followed by the hallux (PrDD) and the first metatarsal head (C1MT). The lowest pressures were registered in the midfoot plant areas. Greater differences between materials were found under the fifth metatarsal head and under the central heel.

Principal components analysis found that pressure distribution variance could be explained by 6 components representing functional and structural aspects:

- General plantar distribution in relation to foot structure,
- Pressures under the heel,
- Pressures under the central midfoot plant,
- Component describing internal heel,
- Component describing toes, first metatarsal head,
- Component describing pressure transmission from the central metatarsal heads to the first toe.

Considering rearfoot movement, the identified principal components described how global variance may be explained by foot position and knee angle at initial contact, as well as ankle inversion and maximum ankle flexion. The role of ankle inversion was considered relevant for transmission regulation and considered to be more human dependent.

With respect to shock absorption, the main findings were in relation to accommodation. Nevertheless, five principal components were identified that describe the variance of shock absorption. Two components were defined by absolute and normalised data with respect to barefoot impact parameters and acceleration. Another two defined a reverse relationship between tibia and upper (head-tibia) transmission, both absolute and normalised and the fifth defined time-to-peak force.

11.6.5.2. Heel pad mechanics

This research provided detailed information about the heel pad mechanics and how it depends on age, gender and obesity which enables the heel pad to act as an effective shock absorber in a variety of circumstances. In relation to passive interaction, the results described in this thesis will promote the development of new solutions for increased shock absorption by the heel pad based on improving rather than restoring its natural shock absorbing characteristics.

The results of this research supported the finding that shock absorption in the lower leg, both barefoot and shod, could be adequately described by a combination of three different mechanical components. However, the meaning of these was not the same in each case demonstrating different heel pad mechanics between shod and barefoot.

The three barefoot heel pad properties components are (i) a component describing elastic deformation under initial loading, (ii) a global elastic component and (iii) a viscoelastic component. The roles of global elastic and initial deformation components are more important in impact mechanics than that of the viscoelastic component. The elastic component, in general, accounted for most of differences in impact forces, while initial deformation component was correlated to the peak-force time. The viscoelastic component showed only a low correlation with the impact force and time-to-peak. The initial deformation component was related to confinement and reorganisation of heel pad under initial loading as a way of increasing deformation without collapse.

Differences in heel pad mechanics due to age, gender and obesity were observed mainly in relation to initial deformation and elastic components. Both seemed to be related to structural human-dependent elements of the heel pad and explained most of the variance in heel pad properties.

Interesting differences were observed in the elderly. The initial deformation component was greater in the elderly than in the young and showed correlation with impact forces whilst the elastic component explained part of the variability of time-to-peak. It is suggested that the initial deformation component was related to reorganisation of the heel pad by means of reshaping in medial-lateral and posterior directions, which could be affected by age degeneration and heel pad confinement. This should be considered in footwear and orthesis design.

A greater elastic component for the non-overweight elderly than for the overweight was, it is suggested, related to compensation of age elasticity loss by increased fat content associated with obesity. Overweight elderly showed lower impact forces and longer time-to-peak than the non-overweight and the young. Peak displacement and time-to-peak were also greater than for the non-overweight and young. Energy absorption and time delays were greater in the overweight. Maximal stiffness, initial stiffness and final stiffness were lower for the overweight than for non-overweight. So, overweight could compensate for age-loss of shock absorbing ability of the heel pad.

Compared with males, females presented a shorter time-to-peak force together with lower energy absorption and time delay and a lower maximal and initial stiffness. The elastic component correlated to time-to-peak in young females but not in young males, probably due to structural differences in heel pad.

These results should be considered in footwear and orthesis design for maintaining and improving the shock absorbing properties of the heel pad, mainly by considering elastic and initial deformation components for adequate heel counter or heel cup design.

Shod heel pad properties are explained in relation to Statement 3. The structure and the materials of the shod heel pad are responsible for its good mechanical properties. Further research should focus on relationship between these and the mechanical components to explain the origin of differences in the general population.

A qualitative model has been proposed comprising three elements equivalents to the three components described elsewhere in this thesis. A mechanical model for initial deformation component (a spring) in series with the viscoelastic and global elastic components arranged in parallel (a spring and dashpot in parallel) could be an appropriate model for overall representation of the heel pad. By changing spring and dashpots characteristics it is possible to include passive interaction, and the shod effect due to confinement and allow for the influence of gender, age and obesity. Modification of initial stiffness by footwear or insole design or by ageing, as shown in this study, would mainly affect the stiffness constant of the first spring whilst human differences would probably affect both springs. This could provide valuable information for footwear designers and serve as a starting point for more sophisticated mechanical or Finite Element Methods based models.

11.6.5.3. Material testing

This thesis has focussed on the reliability of material testing to predict materials' performance in use. It's conclusion is that accurate simulation of the appropriate loading improves the reliability of insert materials testing and the predictability of properties but with respect to upper body transmission, accommodation occurs which gives rise to the differences that are evident between human and material testing. The IBV's method for dynamic testing of materials used in this research yielded good predictions of impact forces and tibia acceleration for insert materials.

Also of interest was the relevant role of **hardness** in impact perception and shock absorption. It is a surface property resulting from simple indentation tests and could be used for material testing as an alternative to more complex machine tests.

An important output of this work was the development of new methods for testing materials by simulating plantar pressures. These methods are part of IBV's commercial services portfolio available to the footwear manufacturers

Dynamic cushioning method showed good results for testing insoles and could be used to predict the ability of materials for reducing plantar pressures.

Recovery-ability measurement. Comfort and heel material perception, as well as rate of loading appear to be related to the recover-ability of materials (Rebote).

Pendulum testing. Heel pad testing is needed to measure passive interaction and for upper body transmission. For these purposes, the instrumented pendulum has proven to be a useful tool for shock absorption analysis

In a general sense, new concepts were identified that suggest that future work should focus on the long term testing of materials, frequency behaviour of materials, human-related properties, initial loading behaviour and surface properties of materials.

11.6.5.4. Passive interaction

Passive interaction is one of the most important results reported in this thesis. It is defined as the mechanical coupling between underfoot materials and heel pad that determine initial elastic deformation and subsequent stiffening under loading. Passive interaction depends on the relative stiffness of insole and heel pad due to their viscoelasticity. If at initial loading the insole was *stiffer* than the heel pad, mechanical coupling will lead to increased contact area, which modifies plantar pressures and thus subjective perception. At the same time, the high residual deformation that would occur in the heel pad would alter its properties. In contrast, if the insole was *softer* than the heel pad, residual deformation would cause insole confinement and modification of heel pad properties. In general, heel pad and insole of similar stiffness would be preferred for optimum heel pad confinement and contact area increase. However, because both heel pad and insole materials are viscoelastic, the stiffness of the system will change with loading frequency to alter the nature of the passive interaction. This concept could be extended to the global interaction between ground surface, foot and insole materials to determine pressures and load perception as stated by Robbins et al. (1988): more reliable estimations of loads with irregular (sharp pressure distribution) than flat ground surfaces

11.7. RECOMMENDED FUTURE WORK

The ultimate purpose of this work was to accelerate the research in shock absorption and this goal has been accomplished. It is intended that the results of this research will be published in appropriate journals. Many, but not all, of the questions raised have been answered and an important list of recommended further work has been identified as appropriate for a range of different biomechanics research areas. New lines of research have been initiated at the Institute of Biomechanics Valencia with projects involving different commercial footwear companies. A new PhD programme to continue and extend this research is in progress at IBV.

Overall, the methodological and experimental work developed as part of this research will provide a reliable foundation for further experimentation.

Recommendations for future work are as follows:

Frequency analysis of shock absorption. Based on the evidence of this research and according to the most recent publications by other authors, impact perception and comfort parameters appear to be best addressed in the frequency domain. Future research in shock absorption should consider the frequency analysis of impact forces and accelerations and their relationship with comfort and impact perception and with material properties. Both, classical Fourier Analysis and time-frequency analysis using wavelets are currently being investigated.

Passive interaction was one of the major results of this work. As a research field passive interaction has different aspects.

- Passive interaction depends on the relative properties of heel pad and insole material and their dependence in frequency. Experiments should be conducted to establish the influence of relative stiffness in both pendulum and walking tests. Changing stiffness with frequency should also be analysed to advance the development of filtering systems.
- In general, heel pad confinement would be preferred to contact area increase. However, the influence of the latter in plantar pressures and perception should be carefully studied. Nonetheless, the pattern of lateral reshaping should be studied together with heel shape to establish an optimum fit to establish design criteria for heel cups. The influence of heel pad confinement has been shown to depend on fitting.
- Heel pad properties play an important role in shock absorption and depend on characteristics of gender, age and obesity. Obesity was observed to benefit elderly people. Further analysis of obesity in the young and in shod people would improve current understanding on heel pad mechanics.
- Material properties at initial loading influence passive interaction. This should be studied further not only in relation to the heel but also for the global analysis of the foot. This could generate a new family of foot friendly materials with improved foot function by enhancing natural systems. The influence in initial loading behaviour of covering materials, internal structures, multilayer or multidensity materials are currently under investigation aiming to generate design criteria for new materials.
- Mass customisation of shoes could become reality through, for example, judicious choice of material and personal heel cup design.

Modelling, as been part of this work and suggest that other models should be explored:

- Both mechanical and Finite Element modelling are considered promising lines of inquiry for new knowledge on impact mechanics and hence the development of innovative solutions to enhance the natural shock absorption properties of the heel pad.
- New work using EQS (structured equations) to identify a statistical model incorporating materials, heel pad, biomechanical and subjective variables is currently being undertaken.

Material testing was an important part of this research and related further work has been identified as follows:

- New material tests developed in the course of this research are already used at IBV in association with the footwear industry. Nonetheless, as discussed in relation to Statement 2, degradation of material properties with use should be further considered and the effect of degraded properties on shock absorption studied. In this context, a fatigue test simulating walking is currently under development at the IBV. There is some evidence that the recovery-ability of materials appear to influence rate of loading and long term subjective opinion, and this concept may enable the prediction of long term performance of materials thus avoiding fatigue testing. This should be investigated by combining fatigue and recovery testing.
- New material tests described in this thesis have used to characterise a limited number of different material families. These results should be extended to develop composite insole materials with material layers of varying chemical composition and structure.
- The existence of an optimum stiffness for insert materials was considered to reduce impact transmission without lowering impact proprioception. Research in this subject would yield criteria for selecting insert materials that would be valuable to biomechanics disciplines such as footwear design, orthopaedics and others.
- Frequency dependent behaviour of materials is an important subject for future study and for the development of new materials. The method developed at the IBV for impact testing that computes parameters in the frequency domain could be a base to devise a series of new tests and parameters to describe the changes in the material's elastic, initial elastic and viscoelastic behaviour with frequency. With

this understanding, materials could be designed to filter the undesired frequencies and assume a specific stiffness for high frequencies loading.

- As a result of this and work elsewhere, a wide range of parameters and concepts are available for every material and the choice of material for a given application is complex. New software tools for multi-option materials choice (expert system) are currently under development at the IBV.
- Hardness played a relevant role in shock absorption and could be considered as a simple and inexpensive method to study materials. However, it is a static parameter describing a global response to indentation and further research is needed before it is adopted.

🗗 **In general**, the methodology used to study impact perception appears to be reliable and further experiments should be conducted to advance the understanding of this phenomenon as follows:

- A time related study of impact perception and comfort comparison would clearly identify disparity between the outcomes of both mechanisms. This would define further and separate experiments for each mechanism, especially in relation to the role of ankle inversion in shock transmission.
- The instrumented pendulum test developed in this research is considered to be a sufficiently reliable method to predict walking results for barefoot and when shod. This method could be used for further studies on heel pad mechanics.

🗗 **Frequency analysis of shock absorption**. As a result of this thesis and agreeing with more recent literature, parameters in the frequency domain seem to better describe impact perception and comfort which are the final goals of humans. In this sense, future research in shock absorption should consider the frequency analysis of impact forces and accelerations and their relationship with comfort and impact perception as well as with material properties. Both, classical Fourier Analysis and time-frequency analysis using wavelets are currently being investigated.

🗗 **Passive interaction** was one of the major results of this thesis. As that, a research field in this aspect has been considered including different aspects.

- Passive interaction depends on heel pad and insole material relative properties and their dependence in frequency. Experiments should be conducted to establish the influence of relative stiffness in both pendulum and walking tests results. On the other hand, change of stiffness with frequency should also be analysed to advance in the development of filtering systems.

- In general, heel pad confinement would be preferred to contact area increase. However, the influence of the latter in plantar pressures and perception should carefully be studied. Nonetheless, the influence of heel pad confinement has been shown to depend on fitting. The pattern of lateral reshaping should be studied together with heel shape to establish an optimum fit. Criteria to design heel cups would be then obtained.

- Heel pad properties play an important role in shock absorption and depend on individual characteristics as gender, age and obesity. Obesity was observed to benefit elderly people. Further analysis of obesity in the young and shod people would improve current understanding on heel pad mechanics.

- Material properties at initial loading influence passive interaction. This should be further studied not only in relation to the heel but also for the global of the foot. As a result a new generation of foot friendly materials could be developed to improve foot function by enhancing natural systems. In this sense, the influence in initial loading behaviour of covering materials, internal structures, multilayer or multidensity materials are currently under investigation to generate design criteria for new materials.

- Mass customisation of shoes could benefit by allocating the best material, heel cup design and so on for each person.

Modelling, as proposed along the thesis and from the results obtained both mechanical and Finite Element modelling are promising lines of work to develop new knowledge on impact mechanics as well as innovative solutions to improve shock absorption by using natural shock absorbers. New work using EQS (Structured equations) to identify a statistical model including material, heel pad, biomechanical and subjective variables is currently being done.

Material testing was an important part of this thesis. In consequence, ideas for future line of work were identified.

- New material tests were developed in this thesis and have been already added to the serviced that the IBV offers to the footwear industry. Nonetheless, as discussed in relation to statement two, evolution of material properties with use should be considered and its influence on shock absorption studied. In this sense, a fatigue test simulating walking has been set up at the IBV and is currently under development. In this sense, recovery ability of materials appeared to influence rate of loading and long term subjective opinion. It could be thought to predict long term performance of materials avoiding fatigue testing. This should be investigated by combining fatigue and recovery testing.

- In any case, material tests developed in this thesis were applied to a limited number of different material families. These results are to be extended to other types of materials differing not only in chemical set up but also showing structures, different material layers and so on.

- The existence of an optimum stiffness for insert materials was considered so to reduce impact transmission without lowering impact proprioception. Research in this subject would yield criteria for selecting insert materials that would be of great help in different fields of biomechanics as footwear design, orthopaedics, etc.

- Frequency behaviour of materials is an important subject for future study and development of new materials. The method developed at the IBV for impact testing that computes parameters in the frequency domain could be used as a base to devise a series of new tests and parameters that describe the evolution of material elastic, initial elastic and viscoelastic behaviour with frequency. Then, materials could be designed to filter the undesired frequencies, show a given stiffness for high frequencies, etc.

- As a result of this thesis, and considering other works, a wide range of parameters and concepts are available for every material. In a sense, this makes difficult to choose a material for a given application. New software tools for multi-choice materials choice are currently under development at the IBV.

- Hardness played a relevant role in shock absorption and could be considered as a simple and inexpensive method to study materials. However, it is a static parameter describing a global response to indentation and further research is needed before widely using it.

In general, the methodology used to study impact perception appeared to be reliable and future experiments should be conducted to advance in impact perception. The study of impact perception in time in comparison to comfort would allow a clear separation between the goal of both mechanical systems identified that finally would allow to define further experiments separately for each system, specially in relation to ankle inversion role in shock transmission.

The instrumented pendulum developed in this thesis appeared to be a reliable method for testing barefoot and shod humans so to predict walking results. This method could be used for further studies on heel pad mechanics.

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APPENDIX

Appendix A1. Ergonomic design of the pendulum

The dimensions of support elements and regulation distances in the pendulum were chosen from athropometrics. Considering testing position, the anthropometric dimensions shown in Figure A1.1 were taken for 5 percentile for children 4 years old as lower limit and 95 percentile for adult males as upper limit to fit the testing device for a great range of population (the data was obtained from a work carried out in 1994 on the Spanish population by the IBV, sponsored by Valencia Government and Spanish Society for Industrial Design Development-DDi; Page, 1992).

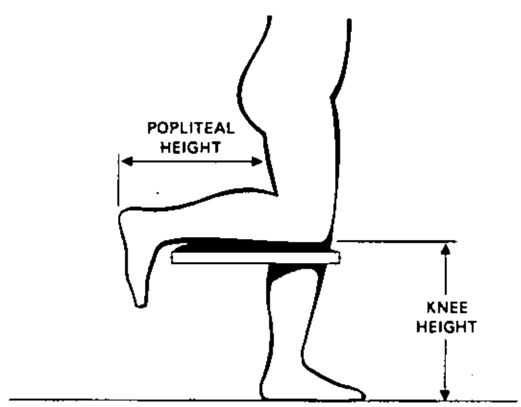


Figure A1.1. Body dimensions considered for pendulum design

The popliteal void (Figure A1.1) was considered to assure good support of the leg. The supported length ranges between 25 cm for small children and 50 cm for adult males. The dimensions of wooden pieces for leg support were also chosen for comfort according to the width of people's leg. The smaller is 160 x 80 x 35 mm and the larger is 160 mm square and 20 mm thick. The length of the rod of the fixation head corresponds to the range of the knee height in the population (Figure A1.1).

Appendix A2. Accelerometer calibration

The experimental set-up used for accelerometer calibration is shown in Figure A2.1. including:

- Accelerometer and electronics used in the pendulum, data card (Acquisition Data Card PCMICIA DAQCard-700 from National Instruments) and a personal computer.
- Precision Goniometer.
- Software developed for Data Acquisition.

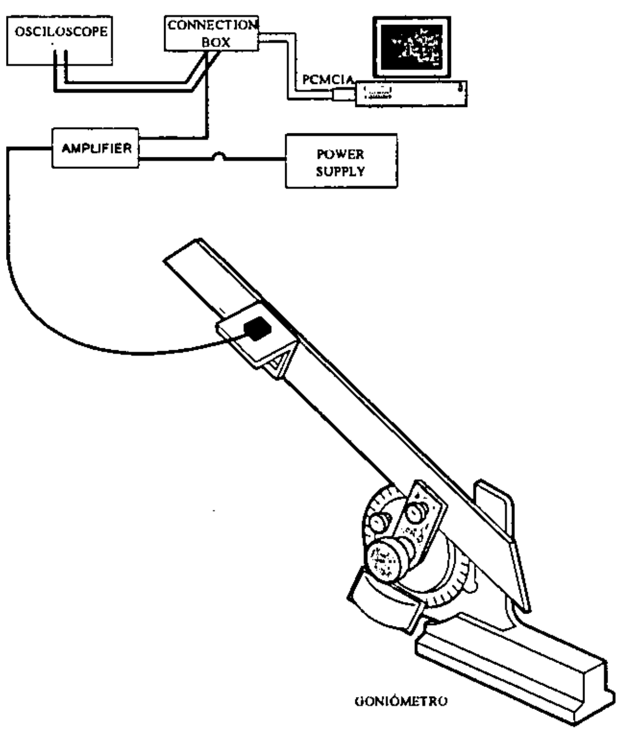


Figure A2.1. Accelerometer celibration set up

The goniometer was disposed in a perfectly horizontal and even surface. It was fixed in this surface with the fix arm perfectly horizontal. Horizontality was assessed using bubble levellers. The accelerometer was mounted in the mobile arm of the goniometer by a double side adhesive tape in such a way that the active axis of the accelerometer was perpendicular to the arm. With this set up the acceleration is 1g for 0 degrees in the goniometer, zero for 90 degrees and -1g for 180 degrees. The acceleration in g's correspond with the cosine of the angle of the mobile arm of the goniometer with respect to the horizontal.

The signal in volts for each goniometer position was acquired in the computer and stored for further calculation. The sampling frequency was adjusted to 100 Hz and 0.5 seconds were registered. Data triggering was set to by keyboard ("por programa" option).

Once all the connections had been checked the equipment was switched on and ten minutes were allowed for warming up. Nine readings were taken between 0 and 180 degrees, signals for 0, 30, 45, 60, 90, 120, 135, 150 and 180 degrees were collected. Degrees and volts were exported to Excel (from Microsoft) where acceleration was calculated for each accelerometer position and linear regression was made for the acceleration and the corresponding value in volts. The resulting equation was implemented in the function conver1.m in the software for data processing. The calibration equation for this accelerometer in m/s2 was:

acel (g's) = 9.81*(1.832*volts-0.0006)

Data collected are included in table A2.1 whereas regression results obtained for pendulum and wall accelerometers are presented in table A2.2.

Teble A2.1 Data collected

Oegrees	Radians	Acceleration (g's)	Volts wall accel.	Volts pendul accel.
0,00E+00	0,00E+00	1,00E+00	5,37E-01	5,27E-01
3,00E+01	5,24E-01	8,66E-01	4,63E-01	4,69E-01
4,50E+01	7,85E-01	7,07E-01	3,79E-01	3,92E-01
6,00E+01	1,05E+00	5,00E-01	2,63E-01	2,83E-01
9,00E+01	1,57E+00	6,13E-17	-7,40E-03	1,51E-02
1,20E+02	2,09E+00	-5,00E-01	-2,78E-01	-2,64E-01
1,35E+02	2,36E+00	-7,07E-01	-3,86E-01	-3,82E-01
1,50E+02	2,62E+00	-8,66E-01	-4,82E-01	-4,75E-01
1,80E+02	3,14E+00	-1,00E+00	-5,43E-01	-5,61E-01

Tabla A2.2. Regression results

	Pendulum accelerometer		Wall accelerometer	
	Coefficients	Standard error	Coefficients	Standard error
Intercepción	-0,00062429	0,00753967	0,01130697	0,00208458
CH2 (TARJ.)	1,83231238	0,01854145	1,84498945	0,00515952

Appendix A3. Acquisition Software operation

The software used for data acquisition is described in this appendix as well as the general operation procedure. This software is a version of a general acquisition programme developed at the IBV.

A3. 1. General operation

When the software is started a work window appears on the screen (Figure A3.1). This window can be divided into three different areas:

1. The graphics area is where the signals acquired are automatically represented versus time. A pointer represented by a cross is available in this area for consulting milivolts and miliseconds corresponding to the point of the signal where it is placed. Graphic representation is intended for checking the quality of the signal before saving.

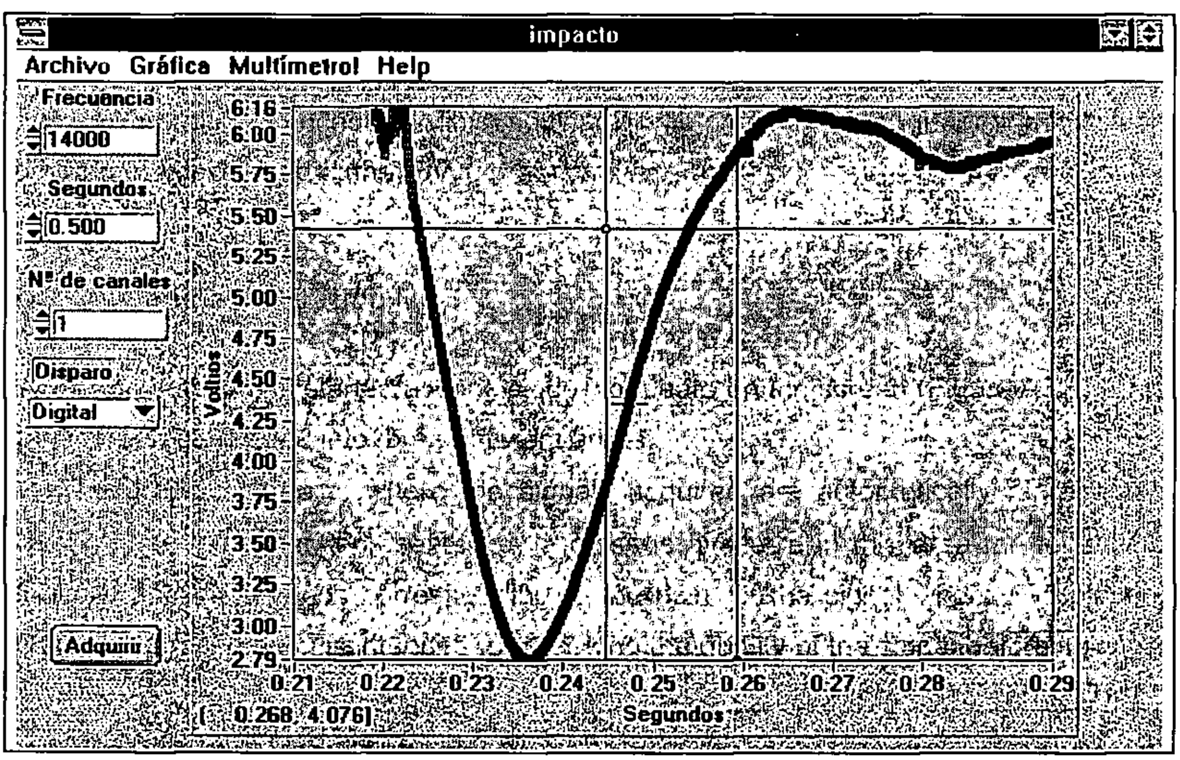


Figure A3.1. Pavi main window

2. The utilities bar, placed in the top includes tools for file and graphics management as well as for testing set up. The following pop-up menus are included:

Archivo. This menu includes the following utilities:

Salir. To quit the programme.

Save. To save the data. Which can be done in two formats: ASCII or in Matlab format.

Configuración. To specify some parameters for data collecting. This utility is set up by the window shown in the Figure A3.2. The following parameters can be specified:

Range (Rango bipolar), sampling range in volts, which is related to resolution. Resolution is 5 mV in the 20 V range.

Reset (Cero previo). Should be off.

G1 to G8. Related to gain of each of the eight channels.

Modo, Nivel (V) and C.Dig. de Disparo, are not used for heel pad testing so they will not be explained here.

Grafica. This menu includes utilities for a better view of the signal. A Zoom utility can be used. It is controlled in the graphics area by two points describing upper left and lower right corners of the diagonal of the zoom window in the graph.

Volver undoes the zoom.

Ayuda. This a help facility about the software.

The utility menu **Multímetro!** is not described here since it is not used for heel pad testing.

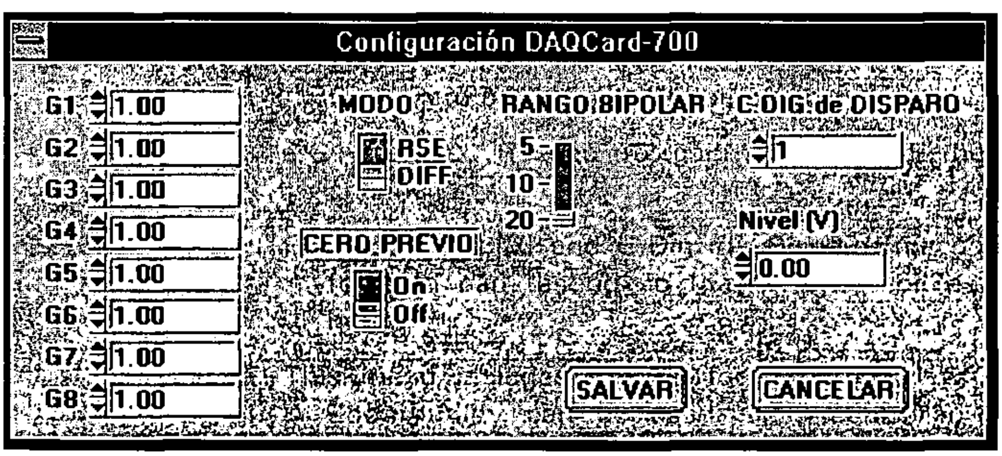


Figure A3. 2. Set up window

3. The control area on the left contains the main control facilities for data collecting.

Frequencia. To specify by keyboard the total sampling frequency. Total means that the frequency input on the box is for all the active channels, i.e. for all the signals to collect. The data card will sequentially collect a datum from each channel with a frequency equal to the total frequency divided by the number of channels. So, a consideration must be done in the number of signals. As an example, if two signals are registered each signal will be acquired with half the frequency typed.

Segundos. To set the time of collection. It has to be born in mind that the larger the data file the slower the data saving and processing. In this sense, the number of samples is equal to the frequency times the time.

N ° de canales. To specify the number of channels (signals) to be acquired. The maximum is eight.

Information of number of samples and coordinates corresponding to pointer position on graphics window are given just down of where time is specified.

Disparo. This is to indicate the way of data triggering, different options are availablc:

Programa to initiate data registering by keyboard.

Digital to begin acquisition by an external signal (photocell for example).

By level. Data acquisition begins once signal has reached a given level.

Adquirir button, to start data acquisition.

A3. 2. Operation procedure

Before testing some set up is required, set up and testing operations are described below.

A3.2.1. Initiating a testing session

Once in the general work window. Several steps must be completed before testing.

1.- Testing set up. Select *configuracion* from the utilities bar and the windows in Figure A5. 3 will appear. The parameters to be specified are as follows.

- Measuring range (In Volts). Range and resolution are directly related. Resolution is limited by Data card's resolution (8 bits). On the other hand, the measuring range has to be wide enough to collect the complete signal without saturation (i.e. the signal would be truncated). For walking tests a range of 10 volts (± 5V) allows to register all the signal with no saturation. The resolution is 25% of measuring range (in volts). It is 5 mv for 20 volts range and 1.25 mv in 5 volts range.

- Set off the previous zero option.
- The rest of controls must be as in the Figure A3. 2 since they need not to be used for heel pad testing.
- Double click Salvar for saving this configuration.

2.- Control set up. Back in the general window. Specify the number of signals to be registered, the global sampling frequency and the time of collection.

3.- Select the type of triggering (*Disparo*). Using a photocell, digital triggering must be selected.

Now the software is ready for testing, these options should not be changed along a testing session.

A3.2.2. Testing

The general testing procedure will be described later. However, some points about the software have to be born in mind.

Once the pendulum has been placed at the drop height and all the instruments are ready impact test can be performed. Double click in the control button (Adquirir) starts acquisition procedure, the computer will await for photocell signal for data triggering, acquisition will finish once elapsed the specified collecting time.

If previous data had not been saved when pressing Adquirir the message "previous data has not been stored, continue? Y/N" would be displayed and the programme would await for confirmation, if "Y" is pressed data on the graphics area will be dismissed and then photocell signal will be awaited for initiating collection. Otherwise, signal will not be registered. This has to be born in mind to avoid unnecessary impacting to the participant because of no acquisition of the signal when performing a test after a non-acceptable trial.

The sampling frequency must be high enough to avoid noise aliasing. It must be at least double of the frequency of the noise, which is 3500 Hz. So the minimum frequency per channel should be 7000 Hz for noise antialising.

Once acquisition has finished, the signal is automatically displayed in the computer screen. For better assessing the impact quality, a graphic zoom is recommended. The corners of the zoom window are defined in the screen with the mouse so to check the desired part of the signal. Double click in zoom in the menu GRAFICA in the utilities bar for inspection and click Volver in the same menu for returning to the prior representation.

A3.2.3. Saving

The programme allows data saving both in ASCII (*.dat) and in MATLAB (*.mat) formats. In both cases data is saved in a matrix with two columns per signal and as many rows as samples. The first column is the time and the second the signal in volts.
Signal processing is performed in a program developed in Matlab. Even though both formats can be used in Matlab, the Matlab format is preferred since storing and processing are faster and requires less storage memory.
After double clicking SAVE, a window will appear allowing to place the files in the desired directory. Write file-name and press Accept. Next saving operation will automatically drive the user to the directory where latter saving was done.
A consideration about the file name must be made when using Matlab format. The name used for creating the file is assigned internally to the data matrix, in case the file name will be changed afterwards, the data matrix will keep responding after the creation name. However, when processing data, Matlab uses matrix name instead of data file name. As a result, if both names are not the same processing will be interrupted by errors in software. In this case, file name should be changed, what would complicate processing. So, to not complicate data processing, it is recommended not to change first data file's name. On the other hand, Names beginning with a number are not valid for data files in Matlab format.

Appendix A4. Software description

Several new testing methods were developed as part of the work conducted in this thesis. At the same time, quite parameters were described and extracted from force and displacement signals obtained from those test. New software was developed to carry out this work. All programmes were developed by the author under Matlab 4.0 for windows using many functions from Matlab libraries.

The software developed in this thesis was:

Pendul, for processing data from pendulum tests. There were two versions as the method was finally improved.

Parama, for parameter calculation from static compression tests.

Cusi, to process and calculate parameters from dynamic cushioning tests.

These programmes were developed as a core programme, which used a series of functions in charge of different common tasks such as filtering, integration, etc. in such a way that the same functions could be used in all the programmes.

The software is described below and the ".m" files are listed at the end of the appendix.

A4.1. Pendul

The software developed for pendulum signal processing is described. This includes two versions for pendulum processing, one with double integration of acceleration used in chapter 7 and the other which was used in chapter 10 when a laser transducer was included for measuring displacement. The core programme was different in both cases, but general functions developed were used in both cases.

The first programme (pendul1.m is explained in first place as it was more general than the other (pendul2.m).

A4.1.1. Pendul1

Pendul1 is the main programme and performs the following operations.

Selection of the data file (*.mat). After executing the software the programme positions the user in the directory of the previous data file. A window for user file selection is showed where only the *.mat files appear.

The data file selected is read in Matlab where the data matrix including time and acceleration in volts is created from the file.

Possible errors in the sign of the signal are corrected and the offset of the signal is eliminated.

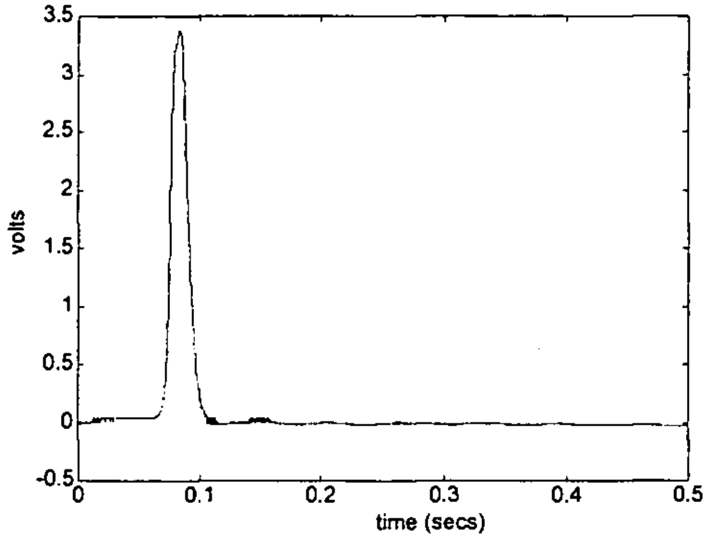


Figure A4.1. Signel from accelerometer (volts).

Signal in volts is showed in the time (Figure A4.1) in a Matlab graphic window where graphical pick is required to restrict the signal to just impact event. User is asked to peak with the mouse at the start and end of impact. The signal is limited to in between the points specified by the user.

Function Corte1 (corte1.m) is called. This function gets the signal restricted by graphical peak and returns the part of the signal corresponding to the impact, which will be processed. A threshold value of 2.5% of peak was used to define start and end of impact. This value can be modified in this function.

Function Filtrar1 (Filtrar1.m) is called. This function applies a two ways low pass digital Butterworth filter to eliminate the effect of vibrations and electrical noise. This function uses the functions butter.m and filtfilt.m from Matlab toolboxes (Matlab, The Math Works, Natick, MA. USA). A cut-off frequency of 150 Hz is used. This frequency has been chosen considering that most of frequency content of heel strike during walking is below 150 Hz (Johnson, 1988; Smeathers, 1989)). A plot of the filtered signal is displayed in Figure A4.2.

Function conver1 (conver1.m) is called. The signal is converted from volts to International units (m/s^2) according to equation resulting from accelerometer calibration. The procedure for calibrating the accelerometer will be later described.

Displacement at the moment of impact is set to zero, that is right since displacement to be studied is that of heel pad.

Function vin (vin.m) is called. This function is used for calculating the impact velocity for integration. In many literature works (Valiant, 1984; Aerts, 1995), impact velocity is calculated from potential energy change. In this work, the velocity at impact is calculated for each test from velocity calibration curves obtained by integrating three acceleration signals from a free oscillation of the pendulum from the testing drop height. The procedure to obtain this curve will be later described in detail. The time passed since data triggering till impact rise is computed. The impact velocity is the value corresponding to that time in the calibration curve velocity.

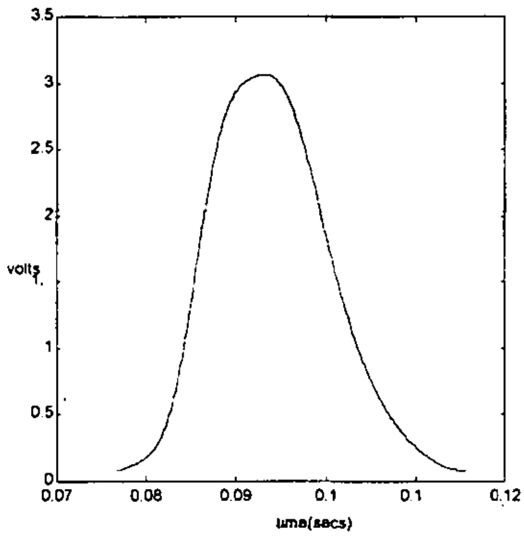


Figure A4.2. Signal after cut and filtering

Function Integra1 (integra1.m) is called. This function performs double integration of the acceleration signal using the initial displacement and the impact velocity as initial conditions. The trapezoidal rule is used for numerical integration. First integration results in a velocity curve (Figure A4.3) and second integration calculates the displacement curve (Figure A4.4). Finally the force is calculated as acceleration times the impacting mass (Figure A4.5). This function returns a matrix with four columns, first is time, second velocity, and third displacement and forth forces. This matrix is used from then for calculating force-displacement (Figure A4.6) and energy absorption parameters.

A more complete description of parameters used for describing heel pad properties was presented in the chapter 7.

Function absor1 (absor1.m) is called. This function calculates the energy absorbed as a percentage of input loading energy:

Energy absorption = (Energy absorbed/Loading Energy) x100

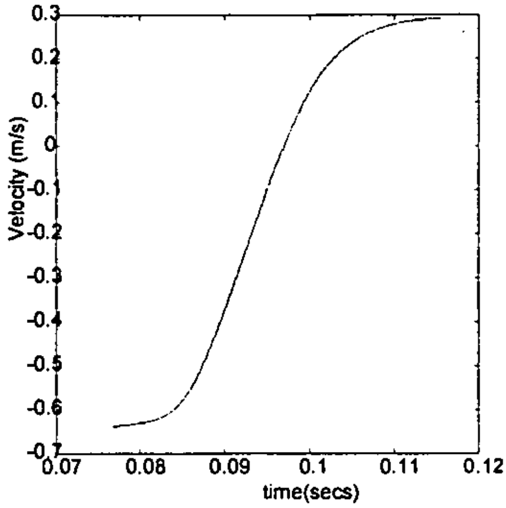


Figure A4.3. Velocity curve

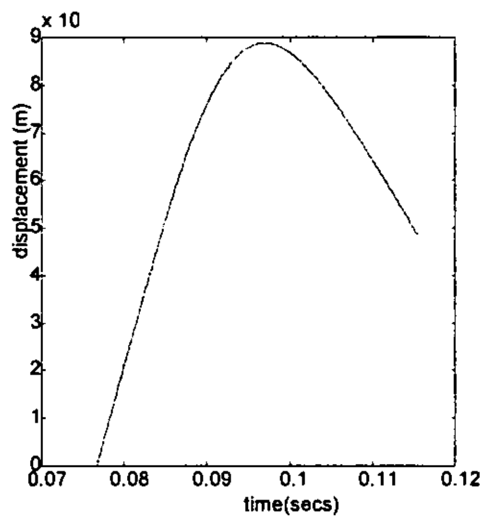


Figure A4.4. Displecement curve

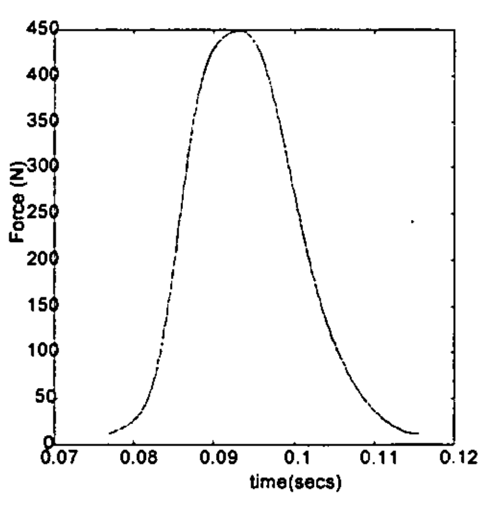


Figure A4.5. Force signal in N.

The energy is calculated as the area under the force-displacement curve. The loading (input) energy is calculated as the area of the loading part of the curve (till maximal displacement) and the return energy is the area under the unloading part. The absorbed energy is the area inside the hysteresis loop curve (Loading Energy - Return Energy) clearly shown in the force-displacement curve (Figure A4.6.). The trapezoidal rule is also used for force-displacement integration.

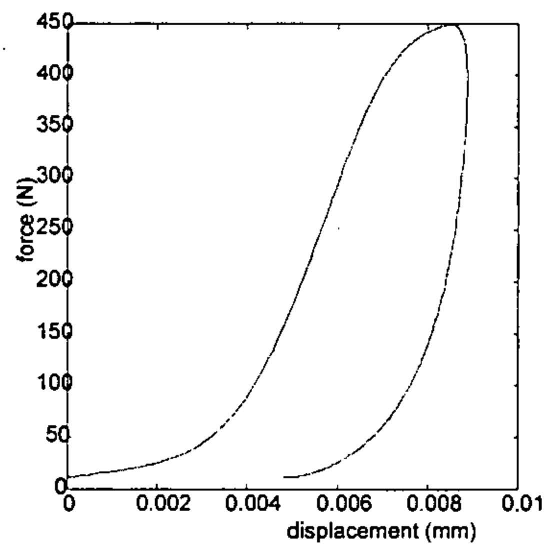


Figure A.4.6. Force-displacement curve obtained by integration

Function pars (pars.m) is called. This function calculates parameters from force-displacement curve. Peak and time to peak are obtained from force and displacement curves using function Max.m from Matlab toolboxes.

Stiffness corresponding at force peak (Kmean) is calculated as peak force divided by the corresponding displacement.

Function rig (rig.m) is called. This function carries out a more complex process to obtain parameters describing the load-displacement curve. As it can be observed in the load-displacement curve and according with previous works, an initial low stiffness region is observed, followed by a high stiffness region and a final medium stiffness till peak force. In this sense, initial (Kini), maximal (Kmax) and final (Kfinal) stiffness for the loading phase as well as the time of shift from initial to maximal stiffness (Tcodo) are calculated (Figure A4.7.). Besides, heel pad displacement continues after maximum force has been reached and a residual displacement is observed at zero level force. Time delay between maximal force and maximal displacement are calculated (Figure A4.7).

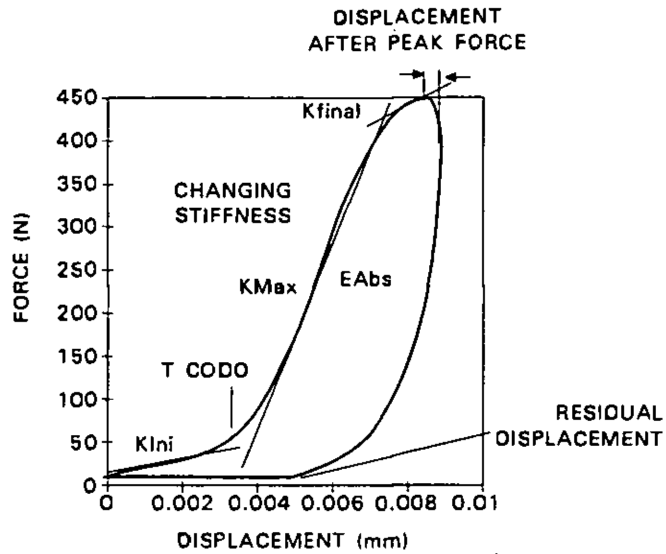


Figure A4.7. Force-displacement paremeters

Function diff.m from Matlab is used in this function to differentiate force with respect to displacement. Diff returns the difference between consecutive values in a column (Xi+1 - Xi). Then the derivative formula is applied: dF/dx = (Fi+1 - Fi)/(Xi+1 - Xi). So, force (F) is derivative against displacement (x) in the loading phase. In this sense, the first derivative (dF/d (x)) represents the stiffness curve as a function of displacement (Figure A4.8). Maximal rigidity is obtained as the maximum of this curve. To calculate the initial and final stiffness, linear regression is used in both regions of the curve. To identify the final of the former and the beginning of the latter, the second derivative with respect displacement is used. This curve indicates change in rigidity as a function of displacement, a maximum in this curve will correspond with maximum change in stiffness (Figure A4.9), i. e. end of initial and beginning of final stiffness. The third derivative is also used to better locate the maximum values as they will be represented by a zero value in the third derivative (Figure A4.10).

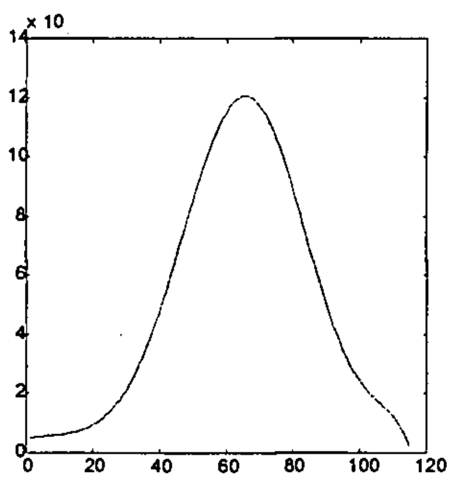


Figure A4.8. First derivetive of Force egeinst displacement till meximum force (stiffness curve).

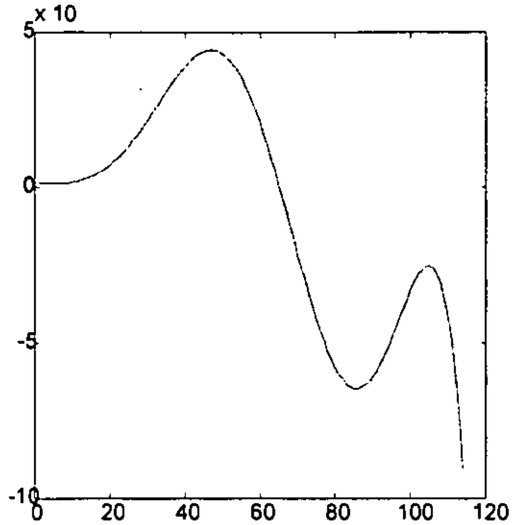


Figure A4.9. Second derivative

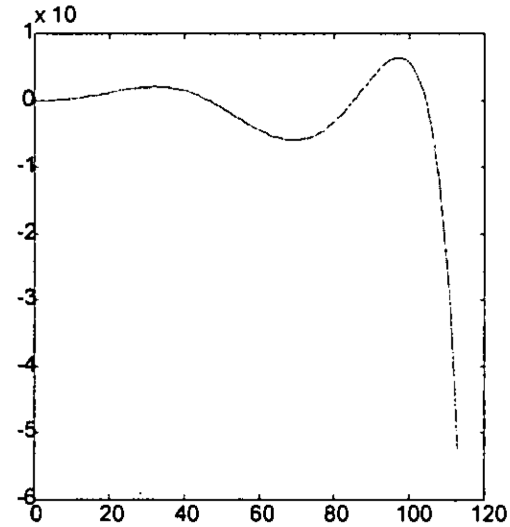


Figure A4.10. Third derivative

All parameters calculated are written in a file (*.par) including the data file name. It is possible to write the parameters obtained from different tests in the same file which can be then directly translated to any statistics package for further analysis.

A4.1.2. Pendul2

In Pendul2 double integration was not performed since displacement was directly measured from the laser transducer. Thus the initial velocity was not needed. It however included the use of interpolation techniques to put all the signals in the same time base.

A4.2. Parama1. Static compression test

This software was developed to process the force -displacement curve obtained from the static compression tests to compute displacement and stiffness at each 50 kPa step.

A4.3. Dynamic cushioning. CUSI

This programme was developed to process force-time, displacement-time and force-displacement signals obtained from each of the ten loading sets used in the dynamic cushioning test.

".M" FILES

```
function pendul

% Software for processing data from pendulum testing.
% Input are *.mat file with two columns of data (tima end acceleration).
% Output are *.par file with parameters describing heel pad properties.
% $$$$$$$$$$$$ FILE NAME READING $$$$$$$$$$$$$$
seguir=1;
while seguir==1
clf
[fichdat,pat]=uigetfile('*.mat',['LECTURA DE FICHEROS (el anterior fué ' fichdat]);

%$$$$$$$$$$$$ DATA FILE READING $$$$$$$$$$$$$$
eval(['load ' fichdat]);
fichdat=lower(fichdat(1:find(fichdat=='.')-1));
A=eval([fichdat]);
eval(['clear ' fichdat]);

%%%%%%%%%% SIGNAL SIGNE %%%%%%%%%%%%%
% Some cases of invertad signals, so signal signe is verified,
% and negative acceleration are corrected.

[mayor instante] = max (abs(A(:,2)));
signo = A(instante);
if (signo < 0 )
    A =-1.*A;
```

```
end

%%%%%%% Graphic cutting of signal %%%%%%%%%
% User is required to specify
% signal beginnign and final
[Ahp thp]=max(A(:,4));
thp=A(thp,3);
[Apa tp]=min(A(:,2));
tp=A(tp,1);
[Apap tpap]=max(A(:,2));
tpap=A(tpap,1);
SF=Apap/Ahp;
TP=tpap-thp;
y=size(A);
plot(A(:,3),A(:,4))
'tica entre los limites'
[a,b]=ginput(2);

%%%     Signal is limited to the impact for further processing
ins=find(A(:,3)>a(1));
inicio=ins(1);
fins=find(A(:,3)>a(2));
final=fins(1);
tcero=A(inicio,3);
tfinal=A(final,3);
A=[A(inicio:final,3),A(inicio:final,4)];

%%%%%%%%%%% Acceleration offset is eliminated %%%%%%%%%
%%%     positive offset is considered as negative.
if (A(1,2)<0)
    A (:,2) = A(:,2) + abs((A(1,2)));
else
    A (:,2) = A(:,2) - abs((A(1,2)));
end
plot(A(:,1),A(:,2))
grid on
'tica entre los limites'
[a,b]=ginput(2);
%%%     limita la señal al pico
ins=find(A(:,1)>a(1));
inicio=ins(1);
fins=find(A(:,1)>a(2));
final=fins(1);
tcero=A(inicio,1);
tfinal=A(final,1);
A=[A(inicio:final,1),A(inicio:final,2)];
if (A(2,2)<0)
    A (:,2) = A(:,2) + abs((A(2,2)));
else
```

```
    A (:,2) = A(:,2) - ebs((A(2,2)));
end

%%%%%%%%%% SIGNAL FILTERING %%%%%%%
% Function filtrar.mis used.
% Cut-off requency (fcorte) is set to 150 Hz.
fcorte=150;
A=filtrar1(A,fcorte);
%%%%%%%%%% PEAK CUTING %%%%%%%%%
% Function corte is used for limiting the signal to be
% only above 2% of maximal acceleration.
A=corte1(A);
plot (A(:,1),A(:,2),'c')
%%%%%%        ACCELERATION IS SET TO INTERNATIONAL UNITS %%%%%
% Function conver1.m is used.
AG = conver1 (A);

%%%%%%%%%%% INTEGRATION %%%%%%%%%%%
% Function integra1.m is used.
% This applications is set for a drop height of 2 cm.
% Initial heel displacement is set to zero.
din=0;

% Impact velocity (m/s) of pendulum is obtained rfo 2 cm drop height
%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%          vinic.m is used. vhdis.mat including velocity curve is required.
%%%%          VELOCITY CHANGE
%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%vin=-0.680;
vin= vinic (AG(1,1))
% INTEGRATION
I = integra1 (AG, vin, din);

%%%%%%%        CALCULATION OF ENERGY ABSORPTION %%%%%%%%%%%%%%%%%%%
% Function absor.m is used, I is the matrix from integra1.m
EABS = absor (I)
%%%%%%%%        CALCULATION OF STIFFNESS FOR PEAK FORCE %%%%%%%%%
[pars tmax] = max(I);
RIG = I(tmax(4),4)/I(tmax(4),3);
[M m] size (I);
defrs=I(M,3);

%%%%%%%%        CALCULATION OF IMPACT PARAMETERS %%%%%%%%%%%%%%%%%%%
Fmax= pars(4);
TFmax=I(tmax(4),1)-I(1,1);
Dmax= pars(3);
TDmax = I(tmax(3),1)-I(1,1);
```

```
pars= rig(l);

%%%%%%%% WRITE OUT PARAMETERS FILE %%%%%%%%%%%
[fichpar,pat]=uigetfile('*.par',['LECTURA DE FICHEROS (el anterior fué ' fichpar]);
if (fichpar==0)
    fichpar=input('Nombre del fichero (entre comillas y sin extension)= ')
    fichpar=[fichpar,'.par']
end

fid2=fopen(fichpar,'a');
fprintf(fid2,fichdat);
fprintf(fid2,'%10.4f ',EABS,RIG,Fmax,TFmax,Dmax,TDmax,pars,vin,tcero, tfinal,SF,TP);
fprintf(fid2,'\n');
fclose(fid2);
seguir=input('¿Deseas seguir (Si=1/No=0)?');
end
end
function parama1
% Este programa calcula parámetros de los ensayos estáticos de materiales
% de calzado
% Los parámetros calculados son:
%                DN (mm), desplazamiento a la presión N (kPa)
%                RN (N/m), rigidez a la presión N. Obtenida por ajuste
%                de los datos a una recta.
%                CRN, Coeficiente de correlación lineal del ejuste
seguir=1;
while seguir==1

%%%%%%%%% LECTURA %%%%%%%%%
[fichdat,pat]=uigetfile('*.asc',['LECTURA DE FICHEROS (el anterior fué ' fichdat]);
if (exist(fichdat)==2)
    [fid1,mensaje]=fopen(fichdat,'r');
    if fid1<2
        mensaje
    end;
    [F]=fscanf(fid1,'%f %1s %f %1s\n',[4,inf]);
    fclose(fid1);
end

% conversión de la fuerza a presión (probeta circular de 50 mm de diámetro)
% la señal está en kg y se pasa primero a N.
% El desplazamiento se convierte a metros
F(1,:)= F(1,:)*9.81;
P=(0.5093*(F(1,:)'));
D=(F(3,:)')*.001;

%%%%%% DIBUJAR Y ELIMINAR TRAMO INICIAL %%%%
yi = find (P>1);
ys = find (P>50);
```

```
li=yi(1);
ls=ys(1);
plot(D(li:ls+10),P(li:ls+10))
title(['Curva presión-deformación para el material ' fichdat])
xlabel('Deformación (m)')
ylabel('Presión (kPa)')
[a,b]=ginput(2);
i=1;
while D(i)<a(1)
    i=i+1;
end
j=1;
while D(j)<a(2)
    j=j+1;
end
C=polyfit(D(i:j),F(1,i:j)',1);
D0=-C(2)/C(1);
D=D-D0;

%%%%% OBTENER PARAMETROS %%%%%
i50=i;
while P(i50)<50
    i50=i50+1;
end
D50=D(i50)
C=polyfit(D(i:i50),F(1,i:i50)',1)
R50=C(1)
NI=0; MI=0;
Pm50=sum(P(i:i50))/length(P(i:i50));
Dm50=sum(D(i:i50))/length(D(i:i50));
for s=i:i50
N=((D(s)-Dm50)^2);
NI=N+NI;
M=((P(s)-Pm50)^2);
MI=M+MI;
end
CR50=R50*sqrt(NI/MI)
i100=i50;
whila P(i100)<100
    i100=i100+1;
end
D100=D(i100)
C=polyfit(D(i50:i100),F(1,i50:i100)',1);
R100=C(1)
NI=0; MI=0;
Pm100=sum(P(i50:i100))/length(P(i50:i100));
Dm100=sum(D(i50:i100))/length(D(i50:i100));
for s=i50:i100
N=((D(s)-Dm100)^2);
```

```
NI=N+NI;
M=((P(s)-Pm100)^2);
MI=M+MI;
end
CR100=R100*sqrt(NI/MI)
i150=i100;
while P(i150)<150
    i150=i150+1;
end
D150=D(i150);
C=polyfit(D(i100:i150),F(1,i100:i150)',1);
R150=C(1)
NI=0; MI=0;
Pm150=sum(P(i100:i150))/length(P(i100:i150));
Dm150=sum(D(i100:i150))/length(D(i100:i150));
for s=i100:i150
N=((D(s)-Dm150)^2);
NI=N+NI;
M=((P(s)-Pm150)^2);
MI=M+MI;
end
CR150=R150*sqrt(NI/MI)
i200=i150;
while P(i200)<200
    i200=i200+1;
end
D200=D(i200);
C=polyfit(D(i150:i200),F(1,i150:i200)',1);
R200=C(1)
NI=0; MI=0;
Pm200=sum(P(i150:i200))/length(P(i150:i200));
Dm200=sum(D(i150:i200))/length(D(i150:i200));
for s=i150:i200
N=((D(s)-Dm200)^2);
NI=N+NI;
M=((P(s)-Pm200)^2);
MI=M+MI;
end
CR200=R200*sqrt(NI/MI)
i250=i200;
while P(i250)<250
    i250=i250+1;
end
D250=D(i250)
C=polyfit(D(i200:i250),F(1,i200:i250)',1)
R250=C(1)
NI=0; MI=0;
Pm250=sum(P(i200:i250))/length(P(i200:i250));
Dm250=sum(D(i200:i250))/length(D(i200:i250));
```

```
for s=i200:i250
N=((D(s)-Dm250)^2);
NI=N+NI;
M=((P(s)-Pm250)^2);
MI=M+MI;
end
CR250=R250*sqrt(NI/MI)
i300=i250;
while P(i300)<300
    i300=i300+1;
end
D300=D(i300)
C=polyfit(D(i250:i300),F(1,i250:i300)',1);
R300=C(1)
NI=0; MI=0;
Pm300=sum(P(i250:i300))/length(P(i250:i300));
Dm300=sum(D(i250:i300))/length(D(i250:i300));
for s=i250:i300
N=((D(s)-Dm300)^2);
NI=N+NI;
M=((P(s)-Pm300)^2);
MI=M+MI;
end
CR300=R300*sqrt(NI/MI)
i350=i300;
while P(i350)<350
    i350=i350+1;
end
D350=D(i350)
C=polyfit(D(i300:i350),F(1,i300:i350)',1);
R350=C(1)
NI=0; MI=0;
Pm350=sum(P(i300:i350))/length(P(i300:i350));
Dm350=sum(D(i300:i350))/length(D(i300:i350));
for s=i300:i350
N=((D(s)-Dm350)^2);
NI=N+NI;
M=((P(s)-Pm350)^2);
MI=M+MI;
end
CR350=R350*sqrt(NI/MI)
i400=i350;
while P(i400)<400
    i400=i400+1;
end
D400=D(i400)
C=polyfit(D(i350:i400),F(1,i350:i400)',1);
R400=C(1)
NI=0; MI=0;
```

```
Pm400=sum(P(i350:i400))/length(P(i350:i400));
Dm400=sum(D(i350:i400))/length(D(i350:i400));
for s=i350:i400
N=((D(s)-Dm400)^2);
NI=N+NI;
M=((P(s)-Pm400)^2);
MI=M+MI;
end
CR400=R400*sqrt(NI/MI)
i450=i400;
while P(i450)<450
    i450=i450+1;
end
D450=D(i450)
C=polyfit(D(i400:i450),F(1,i400:i450)',1);
R450=C(1)
NI=0; MI=0;
Pm450=sum(P(i400:i450))/length(P(i400:i450));
Dm450=sum(D(i400:i450))/length(D(i400:i450));
for s=i400:i450
N=((D(s)-Dm450)^2);
NI=N+NI;
M=((P(s)-Pm450)^2);
MI=M+MI;
end
CR450=R450*sqrt(NI/MI)
i500=i450;
while P(i500)<500
    i500=i500+1;
end
D500=D(i500)
C=polyfit(D(i450:i500),F(1,i450:i500)',1);
R500=C(1)
NI=0; MI=0;
Pm500=sum(P(i450:i500))/length(P(i450:i500));
Dm500=sum(D(i450:i500))/length(D(i450:i500));
for s=i450:i500
N=((D(s)-Dm500)^2);
NI=N+NI;
M=((P(s)-Pm500)^2);
MI=M+MI;
end
CR500=R500*sqrt(NI/MI)
CT=polyfit(D(i50:i500),F(1,i50:i500)',1);
RT=CT(1)
NI=0; MI=0;
Pm=sum(P)/length(P);
Dm=sum(D)/length(D);
for s=1:length(P)
```

```
N=((D(s)-Dm)^2);
NI=N+NI;
M=((P(s)-Pm)^2);
MI=M+MI;
end
CRT=RT*sqrt(NI/MI)
%%%%%% guardar paremetros %%%%%%%%
[fich,path]=uigetfile('*.par',['FICHERO DE PARAMETROS (el anterior fue ' fich]);
if (fich==0)
    fich=input('Nombre del fichero (entre comillas y con extension)= ')
    fid=fopen(fich,'a');
end
fid=fopen(fich,'a');
    fprintf(fid,fichdat);
    fprintf(fid,'
    %f',D0,D50,R50,CR50,D100,R100,CR100,D150,R150,CR150,D200,R200,CR200,D250,R250,CR250,D300
    ,R300,CR300,D350,R350,CR350,D400,R400,CR400,D450,R450,CR450,D500,R500,CR500,RT,CRT);
    fprintf(fid,'\n')
fclose(fid);
seguir=input('¿Deseas seguir (Si=1/No=0)? ');
end
end
```

```
function cusi

% Este programa tratará los datos obtenidos del ensayo denominado cushioning
% de materiales para plantillas.
% La entrada son cuatro columnas: tiempo1, fuerza, tiempo2 y desplazamiento
%
% Las señales de tiempo1 y tiempo2 están desfasados
% Se eliminará el desfase que existe entre las señales
% de fuerza y desplazamiento
%
% Los datos obtenidos de los ensayos se van a filtrar antes  de eliminar el desfase
seguir=1;
while seguir==1

%%%%% PATH %%%%%%%%%%%
p='g:\prj\esport\calzado\cusi\program\dinamic';
path(path,p);
clear p;

%%%%%%%%%% LECTURA DEL FICHERO DE DATOS %%%%%%%%%
[fichdat,path]=uigetfile('*.mat',['LECTURA DE FICHEROS (el anterior fué ' fichdat]);
eval(['load ' fichdat]);
fichdat=lower(fichdat(1:find(fichdat=='.')-1));
T=eval([fichdat]);
eval(['clear ' fichdat]);
% T es la matriz de datos
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
% Conversión de los datos de voltios a fuerza y desplazamiento
% La señal está en voltios y se pasa a kg
% el desplazamiento se pasa e mm
% Cte. 50.97 kg/voltios para pasar a kg
% Cte. 7.59 mm/voltios para pasar a mm
k1=50.97;
k1=k1*9.81; %Se pasa N/volt
k2=7.59;
k2=k2*.001; % se pasa a m/volt

%%% Definición de los vectores F(fuerza) y D(desplazamiento) %%%%
F=T(:,2)*k1;
D=T(:,4)*k2;
t1=T(:,1); %% t1 es el tiempo correspondiente a la fuerza
t2=T(:,3); %% t2 es el tiempo correspondiente al desplazamiento
clear T k1 k2

%%%% Presenta la gráfica de fuerza frente a tiempo
%%%% para eliminar zonas no deseadas
plot (t1,F,'r')
xlabel('Time (mseg)')
ylabel('Force (N)')
title(['Force for material ' fichdat])
% Se ve a preguntar cuantos picos hay
np=input('Número de pulsos=');
% Hay que ticar entre los limites
[a,b]=ginput(np*2);

%%% Cálculo de inicio y final de pulsos %%%
for i=2:2:(np*2);
    ipun=find(t1>a(i-1)); % ipu marca el inicio de pulso
    ipu(i/2)=ipun(1);
    fpun=find(t1>a(i)); % fpu marca el final del pulso
    fpu(i/2)=fpun(1);
end
% en ipu están grabados los tiempos de inicio de pulso
% en fpu están grabados los tiempos de final de pulso
for j=1:np
j
Fx=F(ipu(j):fpu(j)); %% Fx es el vector de fuerza con los datos eliminados
D1=D(ipu(j):fpu(j));
tx=t1(ipu(j):fpu(j)); %% tx es el tiempo correspondiente a Fx
t=t2(ipu(j):fpu(j)); %% t es el tiempo correspondiente a D1
n=size(Fx);

%%%%% Filtra D1 antes de interpolar %%%%%%%%
[coefb,coefa]=butter(4,0.15); %% 0.15=11.25/75 (Frecuencia de corte/la mitad del la frecuencia de muestreo),
    11.25Hz es el llite para una potencia del 98%.
```

```
y=filtfilt(coefb,coefa,Fx);
z=filtfilt(coefb,coefa,D1);
D1x=z;
Fx=y;

%%%%% Subrutina de interpolación para eliminar el desfase %%%%%
% La siguiente función calcula los coeficientes
% de los polinomios que se interpolan
% Sincronizaremos respecto a t1 (tiempo correspondiente a la fuerza)
[c,wkn]=gcvspl(t,D1x,n,ones(n,1),1,3,n,1,1,0,n,6);
[Dx,l,e]=splder(0,3,n,tx,t,c,n);
clear c l e

%%% Dx es el vector de datos eliminando el desfase
%%%%%%%% Cálculo de parámetros %%%%%%%%%%%%%%%%%%%%%%%%%%
I=[tx tx -Dx -Fx];

%% I es una matriz construida para calcular parámetros
Parpur=pars(I);
I=I(Parpur(17):Parpur(18),:);
Parpur=Parpur(1:16);
plot (I(:,3),I(:,4))
xlabel('Displacement (m)')
ylabel('Force(N)')
title(['F-d curve for material ' fichdat])

%%%%%% Guardar parámetros %%%%%%%%%
[fich,path]=uigetfile('*.par',['FICHERO DE PARAMETROS (el anterior fue ' fich]);
if (fich==0)
    fich=input('Nombre del fichero (entre comillas y con extension)= ')
    fid2=fopen(fich,'a');
end
fid2=fopen(fich,'a');
    fprintf(fid2,fichdat);
    fprintf(fid2,' %f',Parpur);
    fprintf(fid2,'\n');
fclose(fid2);
seguir=input('¿Deseas seguir (Si=1/No=0)? ');
end;
end;
end;
```

A4.4. Functions

Functions used for all the programmes are listed below. This includes functions for calculating energy absorption, stiffness from force-displacement curves, filtering, parameters from force and displacement curves and integration.

```
Function B = integra1 (senal, vin, din)

% Integra (signall, vin,din) performs double integration of a signal using trapezoidal rule for numerical
  integration.
% First A is created, A is a matrix N x 2, where N is the signal size.
% vin and din are initial conditions for first and second integrations respectively.
% Outpu is e matrix B(Nx4) wher first column is time from matrix A, second column is first integration, tigrd is
  second integration and fourth is the signal (second column from A).
echo off
format long
% crea A
% La primera columna es el tiempo
% siendo en Labwin T el tiempo muestreado y N el nº de muestras:
% seria matrix1 = [((T/N):(T/N):(T/N)*size(Senal)), ...]
%NN=size(senal);
%Ni=NN(1);
%T=Ni/F;
A=senal;
F=3/(A(6,1)-A(3,1));
[N,n]= size (A);
B(1,1)=A(1,1);
B(1,2)=vin;
B(1,3)=din;
B(1,4)=8.155*A(1,2);
for j=2:N;
    B(j,1)=A(j,1)      ;
    B(j,4)=8.155*A(j,2)         ;
    base=A(j,1)-A((j-1),1);
    delta=abs(base*(A((j-1),2)+((1/2)*(A(j,2)-A((j-1),2)))));
    if A((j),2) < 0;
            delta=-1*delta;
    else delta=delta;
    end
    B(j,2) = B((j-1);2) + delta;
end
for j=2:N;
    base=A(j,1)-A((j-1),1);
    delta=abs(base*(B((j-1),2)+((1/2)*(B(j,2)-B((j-1),2)))));
    if B((j),2) <0;
            delta=-1*delta;
    else delta=delta;
    end
            B(j,3) = (B((j-1),3) + delta);
    end
end
B(:,3)=-1*B(:,3);
B(N,1)= A(N,1);
B(N,4)=8.155*A(N,2);
```

```
plot (B(:,1),B(:,4));
input('pulsa para rigidez')
set(gcf,'Name','RIGIDEZ');
plot (B(:,3),B(:,4));
xlabel('                    desplazamiento');
ylabel('fuerza');
title('rigidez f-d');
end
```

```
function enabs = absor (B)
% This function is used for calculating energy absorption
% Energy is calcualted as percentage of input energy.
% B is the output matrix from integra1
echo off
format long
[N,n]= size (B);
[Y,i]=max(B);
    if i(3)>i(4);
    i=i(3);
    else;
     i=i(4);
    end
area1=0;
area2=0;
for j=2:i;
    base=B(j,3)-B((j-1),3);
    delta=base*(B((j-1),4)+((1/2)*(B(j,4)-B((j-1),4))));
    area1=area1+delta;
    end
for j=i+1:N;
    base=B(j,3)-B((j-1),3);
    delta=base*(B((j-1),4)+((1/2)*(B(j,4)-B((j-1),4))));
    area2=area2+delta;
    end
area=area1+area2;
enabs=(area*100)/area1;
end

%%%%%% comentamos las figuras para mayor claridad en el tratameinto.
%figure (1)
%set(gcf,'Name','RIGIDEZ');
%plot (B(1:i,3),B(1:i,4),'y');
%hold on
%plot (B(i:N,3),B(i:N,4),'b');
%xlabel('                    desplazamiento');
%ylabel('fuerza');
%title('rigidez f-d');
end
```

```
function pars =pars(I)
```

```
%%	Esta función calcula parámetros de las curvas de fuerza
%%	y de desplazamiento.
%%	Devuelve, por orden Area, Fmax, Dmax, Tdmax, Tvisco, Rebotem, enabs, Wdeformación
%%	y Ks(Kfmax (Fmax/d en Fmax), K1 (K máxima), K2 (Kinicial y ordenada en el origen)
%%	t2 (tiempo de paso de K2 a K1), K3 (idem en final)).
%%	Se calculan desplazamiento máximo (Dmax) y tiempo en que ocurre (Tdmax)
%%	Tiempo de diferencia entre el desplazamiento y fuerza máximos
%%	Rebote que es el desplazamiento residual
%%	erea que es el area bajo la curva tiempo-desplazamiento en la fase de descarga
%%	rig calcula los siguientes parámetros:
%% Cálculo de valores máximos (en I la columna 3 es el desplazamiento y la 4 es F).
[par tes]= max (I);
%% Curva de desplazamiento
Dmax= par(3);
Fmax= par(4);
% definir el origen y final de tiempos (al 5% del máximo desplazamiento)
%dinvs=flipud(I(1:tes(3),3));
%t= find (dinvs<(par(3)*.1));
%t0=length(dinvs)-t(1);
%tt=find (I(tes(3):tes(1),3)<(par(3)*.1));
%tf=tt(1)+tes(3);
% idem pero a fuerza menor del 10% del máximo, ya que se trabaja en control de carga
finvs=flipud(I(1:tes(4),4));
t= find (finvs<(par(4)*.1));
t0=length(finvs)-t(1);
tt=find (I(tes(4):tes(1),4)<(par(4)*.1));
tf=tt(1)+tes(4);
Tdmax= I(tes(3),1)-I(t0,1);
Tvisco= I(tes(3),1)-I(tes(4),1);
I=I(t0:tf,:);
clear t tt
Rebote = I(1,3)-I(length(I),3);
%%	Integral del desplazamiento en función del tiempo
%%
area=0;
for j=(tes(3)+1-t0):length(I);
    base=I(j,2)-I((j-1),2);
    delta=base*(I((j-1),3)+((1/2)*(I(j,3)-I((j-1),3))));
    eree=area+delta;
    end
enabs=absor(I);
ks = rig (I);
% rig calcula los siguientes parámetros: Kfmax (Fmax/d en Fmax), K1 (K máxima), K2 (K
% inicial y ordenada en el origen), t2 (tiempo de paso de K2 a K1), K3 (idem en final).
pars =[area Fmax Dmax Tdmax Tvisco Rebote enabs ks t0 tf];
```

```
function rig = rig(I)

%%% Función para calcular rigideces
```

```
%%% tres, inicial, máxima y final
%%% I(Nx4), (t, v, d, f).
%%% deriva por diff.
%%% Calcula K1 (rigidez máxima) K2 (fase  inicial)
%%% t2 (t de fase inicial) y K3 (fase final hasta FMax).
%% limita las señal en la fase de carga y
%% la divide entre hasta Fmax y hasta Dmax
[par tes]=max(I);
II= I(1:tes(4),:);          % Hasta Fmax
Iul=I(tes(4):tes(3),:); % Hasta Dmax
Ides=I(tes(3):length(I),:); % Desde Dmax
%% Primera derivada de F respecto a D (rigidices)
%% hasta Fmax
AFI=diff(II(:,4));
ADI=diff(II(:,3));
%for i=1:length(ADI);
    %if ADI(i)=0;
%   ADI(i)=0.00000000001;
%end
rig1I=AFI./ADI;
%% hasta Dmax.
%AFul=diff(Iul(:,4));
%ADul=diff(Iul(:,3));
%for i=1:length(ADul);
    %if ADul(i)=0;
    %ADul(i)=0.00000000001;
%end
%rig1ul=AFul./ADul;
%% segunda derivada, cambio de la rigidez
AFI2=diff(rig1I);
ADI2=diff(II(1:length(rig1I),3));
rig2I=AFI2./ADI2;
%% tercera derivada
AFI3=diff(rig2I);
ADI3=diff(II(1:length(rig2I),3));
rig3I=AFI3./ADI3;
%% rigidez máxima, de la primera derivada
K1=max(rig1I);
%% rigidez inicial por ajuste hasta el
%% el máximo de la segunda derivada
[PK2 t2]=max(rig2I);
K2=polyfit(II(1:t2,3),II(1:t2,4),1);
t2=I(t2,1)-I(1,1);
%% rigidez final
%% con el minimo de la segunda derivada
[PK4 t4] = max(rig3I);
[PK3 t3]= min(rig2I(1:t4));
K3=polyfit (II(t3:tes(4),3),II(t3:tes(4),4),1);
%% rigidez e fuerza máxima
```

```
%% Fmáxima/desplazamiento en Fmáxime
Kfmax=par(4)/I(tes(4),3);
rig = [Kfmax K1 K2  t2 K3];
%%          Aqui se plotea la curva de rigideces y la de cambios de rigidez
%%          con la de fuerzas y desplazamientos
%%          a fin de poder entender la forma de las curvas y poder
%%          parametrizarlas
%%          Se plotea hasta Fmex y desde Dmax
%%
%rigtot=diff(I(:,4))./diff(I(:,3));
%rmax=max(rigtot);
%rmin=min (rigtot);
%if abs(rmax)>abs(rmin)
%   rr=rmax;
%else
%   rr=-rmin;
%end
```

```
function interpol(action)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%Interpola los datos del ensayo de péndulo a la base de tiempos de la primera columna de
%datos y transforma los datos de voltios a su magnitud física real.
%Lee los nombres de los archivos que van a ser transformados de un fichero de texto y
%guarda en otro archivo con el mismo nombre pero con una 'w' como primer carácter los
%datos en sus unidades correspondientes.
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
if nargin < 1,
  g=figure('name',['TRANSFORMACIÓN DE LOS ARCHIVOS DE PÉNDULO'],'color',[1 1 1],...
       'numbertitle','off','resize','off');
  action='inicio';
end;
%crear menu e iniciar
if strcmp(action,'inicio'),
   nuevo=uimenu(gcf,'label','&Nuevo','callback','interpol(''nuevo'')');
   interpol('nuevo');
end;
%Lee un fichero con el nombre de los ficheros de datos *.mat del ordenador portatil
%donde las columnas son los datos en voltios.
if strcmp(action,'nuevo'),
 [fich,path]=uigetfile('*.txt',['LECTURA DE FICHERO DE FICHEROS']);
 if (fich==0),return;end;
 fid=fopen(fich);
 fichdat=fscanf(fid,'%s',1);
 while fichdat~=[],
   set(gcf,'name',fichdat);
   eval(['load ' fichdat]);
   fname=fichdat(1:findstr(fichdat,'.')-1);
   A=eval([fname]);
     [F C]=size(A);
```

```
 [Y,i]=min(A(:,6));
 li=i-1000;
 ls=i+1100;
 A=A(li:ls,:);
  [F C]=size(A);
 AS(:,1:6)=A(2:F,1:6);
 AS(:,7)=interp1(A(:,7),A(:,8),A(2:F,5));
 AC(:,1)=AS(:,1);
 AC(:,2)=0.603*AS(:,2)+0.0013;
 AC(:,3)=AS(:,3);
 AC(:,4)=-2.018*AS(:,4)-0.0035;
 AC(:,5)=AS(:,5);
 AC(:,6)=8.420*9.8*(-1.435*AS(:,6)+6.044); % Peso del péndulo 8.420 kg.
 AC(:,7)= -2.565*AS(:,7)+12.825;
fname=['w' fname];
 eval([fname '=AC;']);
 clear A AS AC;
 eval(['save w' fichdat ' ' fname]);
eval(['clear ' fname]);
 fichdat=fscanf(fid,'%s',1);
end;
fclose(fid);
end;
return; %FIN FUNCION
```

```
function cortada = corte1 (senal)

% function for reducing signal size.
% signal is restricted to be avor 2.5 % of maximal value.
tt=size(senal);
[Y,i]=max(senal(:,2));
%%%%    corte por delante hasta el 2.5% del pico.
li=find(senal(:,2)>(Y/40));
% li(2) para evitar efectos de espureos
cortada1=[senal(li(2):tt(1),:)];
%%%     corte por detrás
ls=find(cortada1(:,2)<(Y/40));
cortada=[cortada1(1:ls(2),:)];
plot (cortada);
end
```

```
function y=filtrar1(x,fcorte)

% Low passs filter
% Uses butter.m and filtfilt.m from Matlab
% Calculation of sempling frequency
fmuestreo = 3/(x(6,1) -x(3,1));
y(:,1)=x(:,1);
[b,a]=butter(4,2*fcorte/fmuestreo);
y(:,2)=filtfilt(b,a,x(:,2));
```

```
plot(x(:,1), x(:,2), y(:,1), y(:,2),'c');
```

Butter and filtfilt are listed below

BUTTER.M

```
function [num, den, z, p] = butter(n, Wn, ftype, anaflag)

%BUTTER Butterworth digital and analog filter design.
%  [B,A] = BUTTER(N,Wn) designs an N'th order lowpass digital
%  Butterworth filter and returns the filter coefficients in length
%  N+1 vectors B and A. The cut-off frequency Wn must be
%  0.0 < Wn < 1.0, with 1.0 corresponding to half the sample rate.
%  If Wn is a two-element vector, Wn = [W1 W2], BUTTER returns an
%  order 2N bandpass filter with passband W1 < W < W2.
%  [B,A] = BUTTER(N,Wn,'high') designs a highpass filter.
%  [B,A] = BUTTER(N,Wn,'stop') is a bandstop filter if Wn = [W1 W2].
%
%  When used with three left-hand arguments, as in
%  [Z,P,K] = BUTTER(...), the zeros and poles are returned in
%  length N column vectors Z and P, and the gain in scalar K.
%
%  When used with four left-hand arguments, as in
%  [A,B,C,D] = BUTTER(...), state-space matrices are returned.
%
%  BUTTER(N,Wn,'s'), BUTTER(N,Wn,'high','s') and BUTTER(N,Wn,'stop','s')
%  design analog Butterworth filters. In this case, Wn can be bigger
%  than 1.0.
%
%  See also BUTTORD, BESSELF, CHEBY1, CHEBY2, ELLIP, FREQZ and FILTER.

%  Author(s): J.N. Little, 1-14-87
%             J.N. Little, 1-14-88, revised
%             L. Shure, 4-29-88, revised
%             T. Krauss, 3-24-93, revised
%  Copyright (c) 1984-94 by The MathWorks, Inc.
%  $Revision: 1.8 $  $Date: 1994/01/25 17:58:43 $

%  References:
%   [1] T. W. Parks and C. S. Burrus, Digital Filter Design,
%      John Wiley & Sons, 1987, chapter 7, section 7.3.3.

btype = 1;
if (nargin == 3),
    if strcmp(ftype,'s'),
            analog = 1;
    elseif strcmp(ftype,'z')
            analog = 0;
     else   % band-stop or high-pass
            btype = 3;  analog = 0;
    end
elseif (nargin == 4)
```

```
    if strcmp(anaflag,'s'),
            analog = 1;
    elseif strcmp(anaflag,'z')
            analog = 0;
     else
            error('last input ergument should be "s" or "z".');
    end
    btype = 3;  % band-stop or high-pass
else
    analog = 0;
end
if max(size(Wn)) == 2
    btype = btypa + 1;
end
% step 1: gat analog, pra-warped frequencies
if ~analog,
    fs = 2;
    u = 2*fs*tan(pi*Wn/fs);
else
    u = Wn;
end
% step 2: convert to low-pass prototype estimate
if btype == 1             % lowpass
    Wn = u;
elseif btypa == 2         % bandpass
    Bw = u(2) - u(1);
    Wn = sqrt(u(1)*u(2));        % center frequency
elseif btypa == 3         % highpass
    Wn = u;
elseif btype == 4         % bandstop
    Bw = u(2) - u(1);
    Wn = sqrt(u(1)*u(2));        % center frequency
end
% step 3: Get N-th order Buttarworth analog lowpass prototype
[z,p,k] = buttap(n);
% Transform to state-space
[a,b,c,d] = zp2ss(z,p,k);
% step 4: Transform to lowpass, bandpass, highpass, or bandstop of desired Wn
if btype == 1                      % Lowpass
    [a,b,c,d] = lp2lp(a,b,c,d,Wn);
 elseif btype == 2        % Bandpass
    [a,b,c,d] = lp2bp(a,b,c,d,Wn,Bw);
elseif btype == 3         % Highpass
    [a,b,c,d] = lp2hp(a,b,c,d,Wn);
elseif btype == 4         % Bandstop
    [a,b,c,d] = lp2bs(a,b,c,d,Wn,Bw);
end
% step 5: Usa Bilinear transformation to find discrete equivalent:
if ~analog,
```

```
    [a,b,c,d] = bilinear(a,b,c,d,fs);
end
if nargout == 4
    num = a;
    den = b;
    z = c;
    p = d;
else        % nargout <= 3
% Transform to zero-pole-gain and polynomial forms:
    if nargout == 3
            [z,p,k] = ss2zp(a,b,c,d,1);
            num = z;
            den = p;
            z = k;
    else % nargout <= 2
            den = poly(a);
            num = poly(a-b*c)+(d-1)*den;
    end
end
```

FILTFILT.M

```
function y = filtfilt(b,a,x)

%FILTFILT Zero-phase forward and reverse digital filtering.
%  Y = FILTFILT(B, A, X) filters the data in vector X with the
%  filter described by vectors A and B to create the filtered
%  data Y.  The filter is described by the difference equation:
%
%    y(n) = b(1)*x(n) + b(2)*x(n-1) + ... + b(nb+1)*x(n-nb)
%                  - a(2)*y(n-1) - ... - a(na+1)*y(n-na)
%
%  After filtering in the forward direction, the filtered
%  sequence is then reversed and run back through the filter.
%  The resulting sequence has precisely zero-phase distortion
%  and double the filter order.  Care is taken to minimize
%  startup and ending transients by matching initial conditions.
%  The length of the input x must be more than three times
%  the filter order, defined as max(length(b)-1,length(a)-1).
%
%  See also FILTER.
%  Author(s): L. Shure, 5-17-88
%  revised by T. Krauss, 1-21-94
%  initial conditions: Fredrik Gustafsson
%  Copyright (c) 1984-94 by The MathWorks, Inc.
%  $Revision: 1.9 $  $Date: 1994/01/25 17:59:07 $
   error(nargchk(3,3,nargin))
   if (isempty(b)|isempty(a)|isempty(x))
      y = [];
      return
   end
```

```
    [m,n] = size(x);
    if (n>1)&(m>1)
       error('Only works for vector input.')
    end
    if m==1
       x = x(:);  % convert row to column
    end
    len = size(x,1);  % length of input
    b = b(:).';
    a = a(:).';
    nb = length(b);
    na = length(a);
    nfilt = max(nb,na);
    nfact = 3*(nfilt-1);  % length of edge transients
    if (len<=nfact),   % input data too short!
       error('Data must have length more than 3 times filter order.');
    end
 % set up filter's initial conditions to remove dc offset problems at the
 % beginning and end of the sequence
    if nb < nfilt, b(nfilt)=0; end   % zero-pad if necessary
    if na < nfilt, a(nfilt)=0; end
   zi = ( eye(nfilt-1) - [-a(2:nfilt).' [eye(nfilt-2); zeros(1,nfilt-2)]] ) \ ...
       ( b(2:nfilt).' - a(2:nfilt).'*b(1) );
 % Extrapolate beginning and end of data sequence using a "reflection
 % method".  Slopes of original and extrapolated sequences match at
 % the end points.
 % This reduces end effects.
    y = [2*x(1)-x((nfact+1):-1:2);x;2*x(len)-x((len-1):-1:len-nfact)];
 % filter, reverse data, filter again, and reverse data again
    y = filter(b,a,y,[zi*y(1)]);
    y = y(length(y):-1:1);
    y = filter(b,a,y,[zi*y(1)]);
    y = y(length(y):-1:1);
 % remove extrapolated pieces of y
    y([1:nfact len+nfact+(1:nfact)]) = [];
    if m == 1
       y = y.';  % convert back to row if necessary
    end
 function convertg = conver1(signal)
 % (señal) en voltios
 % función para hacer calibrado de la señal pasándola
 % a valores de aceleracion (m/seg2).
 % calibración
 convertg(:,1)=signal(:,1);
 convertg(:,2)=(9.81/.55)*signal(:,2);
 %convertg(:,2) = 9.81*((2.5457471*signal(:,2))-0.03334801);
 end
```

```
 function vinic= vinic(tin)
```

```
% This function calculates impact velocity
% from velocity curve (Vh2)
% amd initial impact time (tin).
% Returns vinic in m/s
% Search the velocity curve (Vh2.mat)
% and loads it for processing.
% This curve is obtained with calivel.m
%load vhdis; %para Vh2
load vhdis05
%para h05
%Vhdis=vhdis;
%para Vh2
%para vh05
%tdis = 1.07228571428571; antes de junio del 96.
%tdis=0.81214285714286; %para vel desde h2
tdis= 0.75614285714286; %para vel desde h05
% Calcula el tiempo inicial de impacto
% sobre la curva de velocidad (tvin).
tin=tin+tdis;
tvin=(find (vhdis(:,1)>tin));
tvin=tvin(1);
% Velocidad de impacto
vinic=vhdis(tvin,2);
```

```
function impuls = impuls (senal)
% Integra la señal f-t para calcular el impulso
% crea primero A, A es una matriz de N x 2, time - fuerza
% Integra la segunda columna respecto de la primera.
% En un principio toma como t de intérvalo el ti de la primera columna.
echo off
format long
A=senal;
[N,n]= size (A);
impuls=0;
for j=2:N;
    base=A(j,1)-A((j-1),1);
    delta=abs(base*(A((j-1),4)+((1/2)*(A(j,4)-A((j-1),4)))));
    if A((j),4) < 0;
            delta=-1*delta;
    else delta=delta;
    end
    impuls = impuls + delta;
end
```

A4.5. Frequency Analysis

The function sdof2.m was developed to compute natural frequency and damping coefficient of pendulum during free oscillation.

```
function [df,fn,fa]=sdof2(dat,t,n)

%[df,fn,fa]=sdof2(dat,t,n)
%  Esta función devuelve el factor de amortiguamiento
%  (df),la frecuencia natural de la oscilación(fn) y
%  la frecuencia amortiguada (fa), de la oscilación dat
%  referida al eje de tiempos t.
%  n: Número de máximos que calcula. Para cada par calcula
%     los parametros df,fn,fa.
%@IBV 1995 Marta Mateu & Arturo Forner
l=length(t);
%He hecho una modificacion para que ajuste bien los valores de los ejes
fs=1/(t(2)-t(1))
plot(t,dat);
hold on
[x,y]=ginput(2*n);
for i=2:2:2*n,
    t1=floor(x(i-1)*fs);
    t2=ceil((x(i)*fs)+1);
    [ampli,pos]=max(dat(t1:t2))
    pos=t1+pos;
    a(i/2)=ampli;       %Guarda las amplitudes máximas
    b(i/2)=pos;         %Guarda el instante en que se producen las amplitudes máximas
plot(t(b),a,'xr');
end;
for j=1:(n-1),
    de=log(a(j)/a(j+1));
    z=(de/(2*pi))^2;
    df(j)=sqrt(z/(1+z));
    fa(j)=1/(t(b(j+1))-t(b(j)));
    fn(j)=fa(j)/sqrt(1-df(j)^2);
end;
end;
```

Appendix A5. Frequency Analysis of a pendulum impact

Fast Fourier Transform (FFT) of a signal was done in MatLab (Figure A8.1) using a square window. It was observed, as expected, that most of the signal was under 150 Hz, so this frequency could be used for low pass filtering in data processing, finally a cut-off frequency of 100 Hz was selected. Spectral power analysis revealed that less than 2% of information was loss using this filtering frequency.

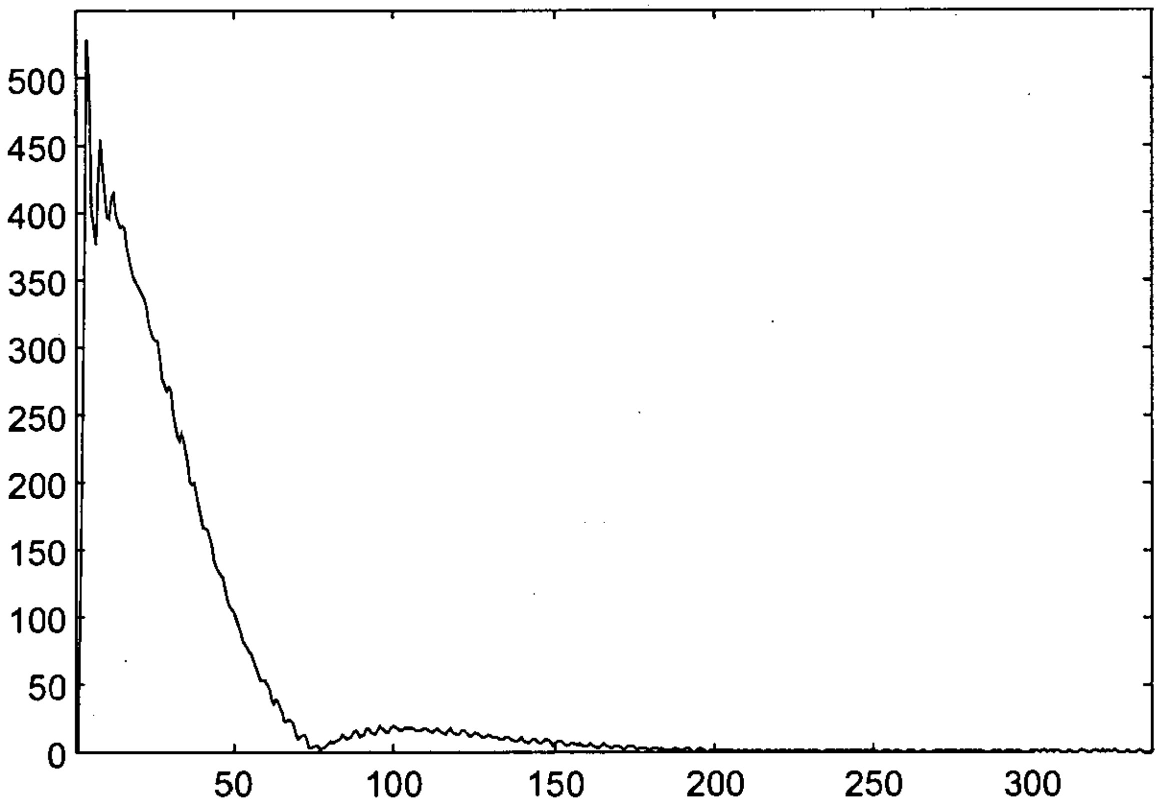


Figure A5.1. Frequency spectrum of an impact

Appendix A6. Questionnaire

FINAL COMFORT TEST

Sports Biomechanics Group. Footwear Section. JULY/1999

This questionnaire is CONFIDENTIAL. The data in it will be part of a larger study. No reference will be made to a particular questionnaire.

IMPORTANT

This questionnaire is about footwear you have been wearing longer than an hour, and you are still wearing when answering this questionnaire.

NAME: **CODSUJ:**

CODZAP: **CONFORT: C____**

DATE: ....

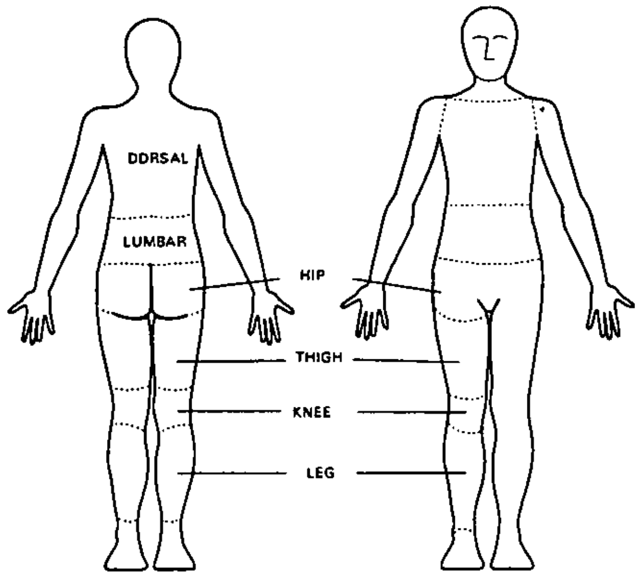
COMFORT TEST IN BODY AREAS

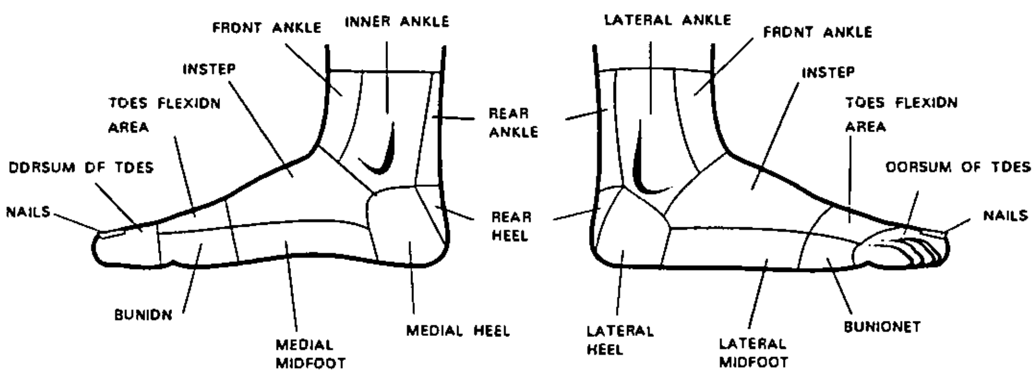
THE PARTICIPANT MUST HAVE THE BODY AREAS PICTURE IN FRONT OF HIM/HER

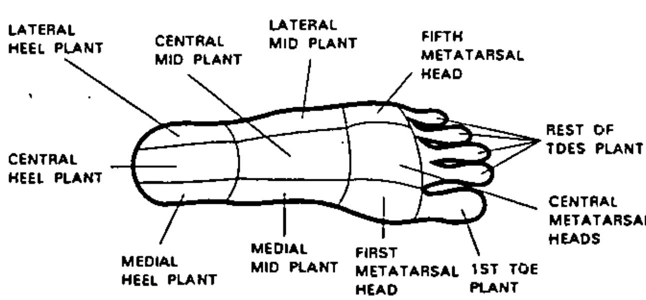
Please indicate the level of discomfort or pain you are feeling AT THE MOMENT and at each body area BEFORE TAKING the shoes OFF. According to the following scale, circle the area, side (right or left) and level of discomfort in the table. (**D**: Right; **I**: Left)

		0	1	2	3	4	S	6
	AREAS	**No dis-comfort**	**A bit of discomfrt**	**Moderate discomfrt**	**Severe discomfrt**	**A bit of pain**	**Pain**	**Severe pain**
MFDDRS	Dorsal area	D I	D I	D I	D I	D I	D I	D I
MFLUMB	Lumbar area	D I	D I	D I	D I	D I	D I	D I
MFCADA	Front hip	D I	D I	D I	D I	D I	D I	D I
MFCADP	Rear hip	D I	D I	D I	D I	D I	D I	D I
MFMUSA	Front thigh	D I	D I	D I	D I	D I	D I	D I
MFMUSP	Rear thigh	D I	D I	D I	D I	D I	D I	D I
MFRODA	Front knee	D I	D I	D I	D I	D I	D I	D I
MFRODP	Rear knee	D I	D I	D I	D I	D I	D I	D I
MFPIA	Front leg	D I	D I	D I	D I	D I	D I	D I
MFPIP	Rear leg	D I	D I	D I	D I	D I	D I	D I
MFTOBA	Front ankle	D I	D I	D I	D I	D I	D I	D I
MFTOBP	Rear ankle	D I	D I	D I	D I	D I	D I	D I
MFTOBI	Inner ankle (tibial)	D I	D I	D I	D I	D I	D I	D I
MFTOBE	Extl. ankle (peroneum)	D I	D I	D I	D I	D I	D I	D I
MFTALP	Rear heel	D I	D I	D I	D I	D I	D I	D I
MFTALI	Inner heel (tibial)	D I	D I	D I	D I	D I	D I	D I
MFTALE	Extl. heel (peroneum)	D I	D I	D I	D I	D I	D I	D I
MFEMPE	Instep	D I	D I	D I	D I	D I	D I	D I
MFFLDD	Toe flexion area	D I	D I	D I	D I	D I	D I	D I
MFDSDD	Dorsum of toes	D I	D I	D I	D I	D I	D I	D I
MFUÑAS	Nails	D I	D I	D I	D I	D I	D I	D I
MFMPE	Lateral midfoot	D I	D I	D I	D I	D I	D I	D I
MYMPI	Medial midfoot	D I	D I	D I	D I	D I	D I	D I
MFJUAN	Bunion	D I	D I	D I	D I	D I	D I	D I
MYJLLO	Bunionet	D I	D I	D I	D I	D I	D I	D I
MFPLTI	Medial heel plant	D I	D I	D I	D I	D I	D I	D I
MFPLTC	Central heel plant	D I	D I	D I	D I	D I	D I	D I
MFPLTE	Lateral heel plant	D I	D I	D I	D I	D I	D I	D I
MFPLMI	Medial mid plant	D I	D I	D I	D I	D I	D I	D I
MFPLMC	Central mid plant	D I	D I	D I	D I	D I	D I	D I
MFPLME	Leteral mid plant	D I	D I	D I	D I	D I	D I	D I
MFC1M	1st metatarsal head	D I	D I	D I	D I	D I	D I	D I
MFCMM	Central metatrsl. head	D I	D I	D I	D I	D I	D I	D I
MFC5M	5th metatarsal head	D I	D I	D I	D I	D I	D I	D I
MFPL1D	1st toe plant	D I	D I	D I	D I	D I	D I	D I
MFPLRD	Plant of other toes	D I	D I	D I	D I	D I	D I	D I

Indicate in the following picture (page 4), if the footwear tested presses your feet (O) or has caused sores (R) at any point of the feet.


DORSAL
LUMBAR
HIP
THIGH
KNEE
LEG


FRONT ANKLE
INNER ANKLE
INSTEP
TOES FLEXION AREA
DORSUM OF TOES
NAILS
REAR ANKLE
REAR HEEL
BUNION
MEDIAL MIDFOOT
MEDIAL HEEL
LATERAL ANKLE
FRONT ANKLE
INSTEP
TOES FLEXION AREA
DORSUM OF TOES
NAILS
LATERAL HEEL
LATERAL MIDFOOT
BUNIONET


LATERAL HEEL PLANT
CENTRAL MID PLANT
LATERAL MID PLANT
FIFTH METATARSAL HEAD
REST OF TOES PLANT
CENTRAL HEEL PLANT
CENTRAL METATARSAL HEADS
MEDIAL HEEL PLANT
MEDIAL MID PLANT
FIRST METATARSAL HEAD
1ST TOE PLANT

SUBJECTIVE PERCEPTION

1. GENERAL

1.1. What do you think of the STABILITY of the shoes (have you noticed that the footwear moves sideways when walking or standing)? JSESTAB

☐ Very stable ☐ Stable ☐ Nor stable neither unstable ☐ Unstable ☐ Quite unstable

1.2. Do you think the shoes are heavy? JSPESO

☐ Very light ☐ Light ☐ Nor light neither heavy ☐ Heavy ☐ Quite heavy

1.3. Regarding weight, how would you prefer the shoes to be? JSPRPES

☐ Lighter ☐ Nor lighter neither heavier ☐ Heavier

1.4. What do you think of the ease of use of the fastening? JSUSOCI

☐ Very easy ☐ Easy ☐ Nor easy neither difficult ☐ Difficult ☐ Quite difficult

2. HEEL

2.1. What do you think of the heel width in contact with the ground? JSTCA

☐ Too narrow ☐ Narrow ☐ Nor narrow neither wide ☐ Wide ☐ Too wide

2.2. How would you prefer the heel width in contact with the ground to be? JSTCPR

☐ Narrower ☐ Nor narrower neither wider ☐ Wider

2.3. What do you think of the heel height? JSTCH

☐ Too high ☐ High ☐ Nor high neither low ☐ Low ☐ Too low

2.4. How would you prefer the heel height to be? JSTCPR

☐ Lower ☐ Nor lower neither higher ☐ Higher

2.5. What do you think of the heel material? JSTCM

☐ Too soft ☐ Soft ☐ Nor soft neither hard ☐ Hard ☐ Too hard

2.6. How would you prefer the heel material to be?

☐ Softer ☐ Nor softer neither harder ☐ Harder

3. SOLE

3.1. When the foot is on the floor and you raise the heel, what do you think of the sole flexibility? JSSUFL

☐ Too rigid ☐ Rigid ☐ Nor rigid neither flexible ☐ Flexible ☐ Too flexible

3.2. What would you prefer regarding the sandal FLEXIBILITY?

☐ Less flexibility ☐ It is OK ☐ More flexibility

3.3. What do you think of the height of the front sole? JSSUHP

☐ Too low ☐ Low ☐ Nor low neither high ☐ High ☐ Too high

3.4. How would you prefer the front sole?

☐ Lower ☐ It is OK ☐ Higher

3.5. What do you think of the insole material at the point? JSSUMP

☐ Too soft ☐ Soft ☐ Nor soft neither hard ☐ Hard ☐ Too hard

3.6. How would you prefer the insole material at the point to be? JSSUPRM

☐ Softer ☐ It is OK ☐ Harder

3.7. What do you think of the insole material at the heel? JSSUMP

☐ Too soft ☐ Soft ☐ Nor soft neither hard ☐ Hard ☐ Too hard

3.8. How would you prefer the insole material at the heel to be? JSSUPRM

☐ Softer ☐ It is OK ☐ Harder

4. FITTING

4.1. What do you think of the sandal length (longitudinal room for the foot)? JSAJL

☐ Too short ☐ Short ☐ Nor short neither long ☐ Long ☐ Too long

4.2. How would you prefer the length of the sandal to be? JSAJPRL

☐ Shorter ☐ It is OK ☐ Longer

4.3. What do you think of the sandal width (transversal room for the foot)?

☐ Too narrow ☐ Narrow ☐ Nor narrow neither wide ☐ Wide ☐ Too wide JSAJA

4.4. How would you prefer the length of the sandal to be? JSAJPRA

☐ Wider ☐ It is OK ☐ Narrower

4.5. Regarding your size, how would you prefer the sandal to be? JSAJPF

1 ☐ One size smaller *2* ☐ It's OK *3* ☐ One size larger

5. GENERAL INSOLE

5.1. What do you think of the insole material? JSPLM_

☐ Too soft ☐ Soft ☐ Nor soft neither hard ☐ Hard ☐ Too hard

5.2. How would you prefer the insole material to be? JSPLPRM_

☐ Softer ☐ It is OK ☐ Harder

5.3. What do you think of the comfort provided by the insole? JSPLCO

1 ☐ Extremely comfortable

2 ☐ Very comfortable

3 ☐ Quite comfortable

4 ☐ Nor comfortable neither uncomfortable

5 ☐ Very uncomfortable

6 ☐ Quite uncomfortable

7 ☐ Extremely uncomfortable

6. WHAT DO YOU THINK OF THE SHOES YOU TESTED?

1 ☐ Extremely comfortable

2 ☐ Quite comfortable

3 ☐ Very comfortable

4 ☐ Normal (nor comfortable neither uncomfortable)

5 ☐ Very uncomfortable

6 ☐ Quite uncomfortable

7 ☐ Extremely uncomfortable

7. DO YOU HAVE ANY COMMENTS ON THE SHOES? FCOMENT

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....

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Appendix A7. Laser transducer details and calibration procedure

Transducer characteristics

Measuring range:	± 10 mm
Focal length:	40 mm
Resolution	40 μm
Linearity:	1% scale
Response time:	0.15 ms
Offset adjustmen range:	± 10 mm
Span regulation Range	0.4 V/mm ± 30 mm
Analogic output:	-4 a 4 V (30* 50 mm)
Maana supply:	12 a 24 Vc.c. ± 10 %, rizado (p-p): 10 % max
Uptake:	120 mA máx.

For laser calibration the procedure followed included de set up presented in the figure A7.1 and included:

Displacement laser transducer

Electronics for signal amplifications and supply

Portable computer with acquisition software

Gauge (accuracy: 0.01 mm)

Mechanical part for mounting the transducer onto the gauge

Verification Marble (horizontal precision surface)

The laser was mounted on the gauge which was placed over the marble in such a way that laser surface was parallel to the marble. Once the laser in place it was connected to the electronics and to the computer as the acquisition software was started.

The laser was placed 50 mm away from the marble surface (maximum measuring range of the sensor). At this distance, the position leds of the sensor, which indicate that it was out of range, should blink. The sensor was then moved from this distance towards the marble registering the analogic output every 2 mm as measured with the gauge till the lower measuring range was reached (30 mm away from the marble). At this moment the laser leds must blink again. By this procedure 10 distance measurements and the corresponding electric signal were obtained. Regression statistics techniques were used then to obtain the calibration curve.

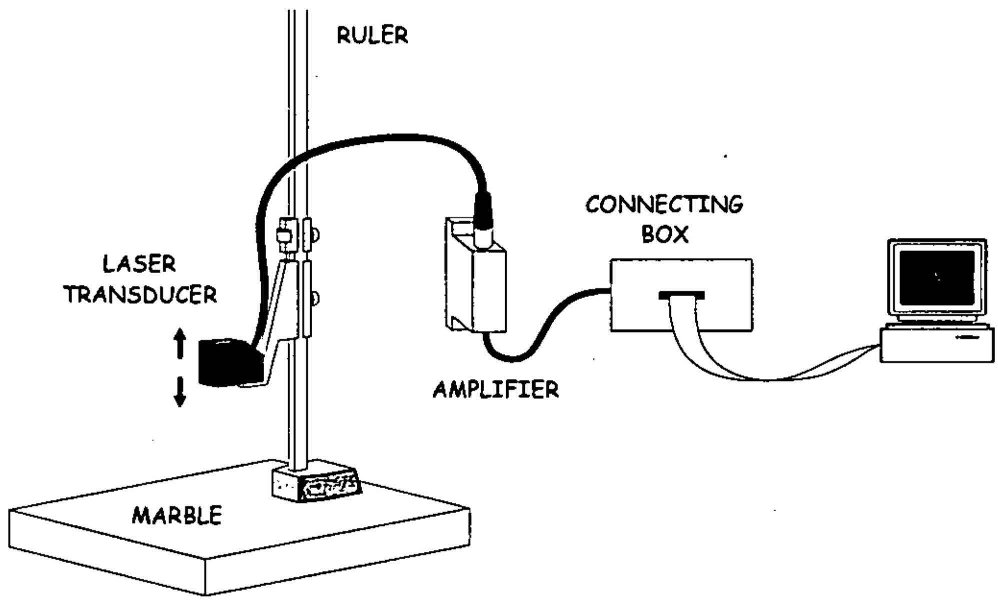

Figure A7.1. Set up for laser calibration