CHARLES UNIVERSITY IN PRAGUE

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DEPARTMENT OF PHYSIOTHERAPY

Detection methods of foot shape and pressure distribution

(Critical review)

Diploma Thesis

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Declaration

I declare that this diploma thesis is written by me with the help of the litterature which I refer to in the end.

A. Andreou

Acknowledgement

I would like to thank my supervisor PaedDr. Karel Jelen, Csc for the assistance and supervision during the realization of this work. I would also like to thank my family and my friends for their help and support during the writing of this thesis.

Abstract

Title of thesis work:

Detection methods of foot shape and pressure distribution – critical review.

Objectives of thesis work: The objective of this thesis work is to analyze the aspects of foot shape and pressure distribution and to describe the various factors which are responsible for measuring foot shape and pressure distribution and tries to describe the different methods to measure the same.

Method: The solution over the objective mentioned above is to review the existing available literature and knowledge about the related topic, summarise it and come into a conclusion.

Results: In this thesis work are described the foot shape and pressure distribution in static and dynamic loading. Also various factors wich affect the foot pressure are analysed and different kinds of methods of how to detect the foot shape and pressure distribution are recorded.

Key words: foot, plantar pressure, gait, standing, detection methods.

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1. Preface

"The human foot is a masterpiece of engineering and a work of art"
(Leonardo Da Vinci)

The foot is one of the most weight- bearing and shock absorbing structure in the human body. Biomechanical factors play an important role on the etiology, treatment and prevention of many foot disorders and gait disturbances.

The significant differences of foot shape and force distribution between individuals, indicates how unique is the human foot and that may contain valuable informations about a patient's gait and his/her foot structure.

It would be very interesting if by anymeans somebody could encrypt these informations and use them for earlier diagnosis, prevention or therapy.

2. Aim of diploma thesis

The aim of this research is to find methods of how to measure (detect) the pressure under the foot, during barefoot standing and walking and methods of how to evaluate the shape of the foot. The research also analyses many aspects of foot shape and pressure distribution and factors which affect it. Finally the research describes various factors which are responsible for measuring foot shape and pressure distribution and tries to describe the different methods to measure the same.

3. Methodology

For the achievement of the goal mentioned above, review of the existing litterature through various databases and books was found to be the best solution in my case. The research took place in first medical faculty library in Prague, the library of faculty of physical education and sport in Prague and through my personal internet during the period 2007 to 2008.

Keyword that were searced include: foot, plantar pressure, gait, standing, detection methods, foot force, gait analysis, foot pressure, foot shape, biomechanics of foot, stress distribution.

4. Gait cycle and its phases

Walking is the main form of animal locomotion on land, distinguished from running and crawling. Gait is a sequence of foot movements by which human is able to transport it self. The word walking is derived from the Old English walkan which means (to roll).

Walking is generally distinguished from running in that only one foot at a time leaves contact with the ground.

Many events are happening at the same time, during walking and it can seem overwhelming. However, the process of walking can be broken down into a series of steps which can go some distance in simplifying the process.

The gait cycle of each leg is divided into the **stance phase and the swing phase**. The stance phase is the period of time during which the foot is in contact with the ground. The swing phase is the period of time in which the foot is off the ground and swinging forward. In walking, the stance phase comprises approximately **60%** of the gait cycle and the swing phase about **40%.** The proportion of swing to stance phase changes as the speed of walking or running **increases.** As the speed is increased the percentage of time spent in the stance phase **decreases.**

4.1. Duration of gait cycle

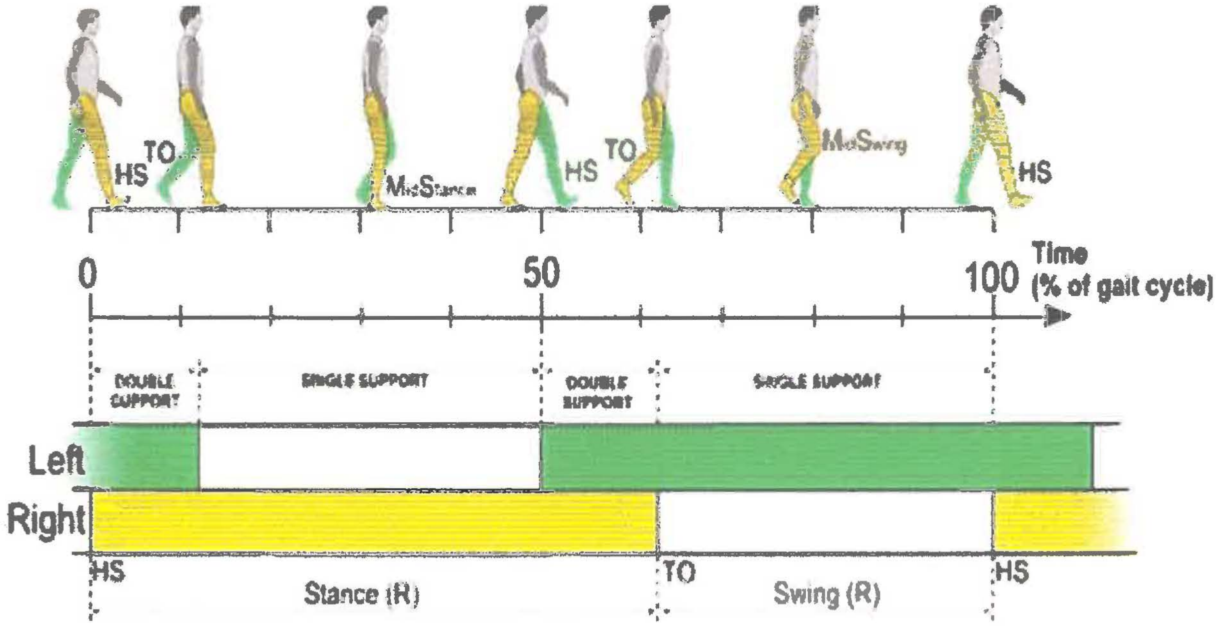

Figure 1. Gait cycle. Obtained from:
(http://www.momentumsports.co.uk/media/images/GaitCycleDiag.jpg)

The above figure shows the steps of the gait cycle and the duration of each one. The duration of initial contact is 27% of the stance phase. The duration of the midstance (loading responce) is the 40% of the stance phase and the terminal stance and preswing is the 33% of the stance phase.

4.2. Gait cycle

The two main phases of gate mentioned above can be divided in four and three sub-phases respectively.

The stance phase is divided to:

- Initial Contact
- Mid-stance
- Terminal stance
- Toe off

The swing phase is divided to:

- Initial swing
- Mid-swing
- Terminal swing

An important point to note is that in running an added sub-phase is present. Float phase. During float phase, neither foot is on the ground.

4.2.1. Stance phase - Initial Contact:

- Is the moment when the red foot just touches the ground
- Is when the heel is the first bone of thee foot to touch the ground
- Meanwhile, the blue leg is at the end of terminal stance
- Shoulder is extended
- Pelvis is rotated left
- Hip is flexed and externally rotated
- Knee is fully extended
- Ankle is dorsiflexed
- Foot is supinated
- Toes are slightly extended

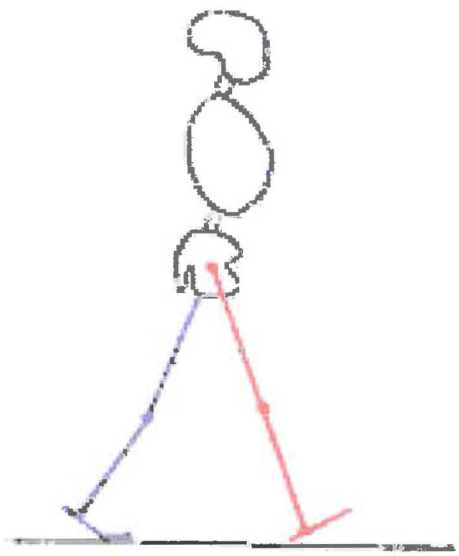

Figure 2. Stance phase, initial Contact. Obtained from: (Kaczmarska A. 2006)

4.2.2. Stance phase. (Loading response) mid-stance:

- The double stance period beginning
- Body weight is transfered onto the red leg
- Phase 2 is important for shock absorption, weight-bearing, and forward progression
- The blue leg is in the pre-swing phase
- Shoulder is slightly extended
- Pelvis is rotated left
- The hip is flexed and slightly externally rotated
- The knee is slightly flexed
- The ankle is plantarflexing to neutral
- Foot is neutral
- Toes are neutral

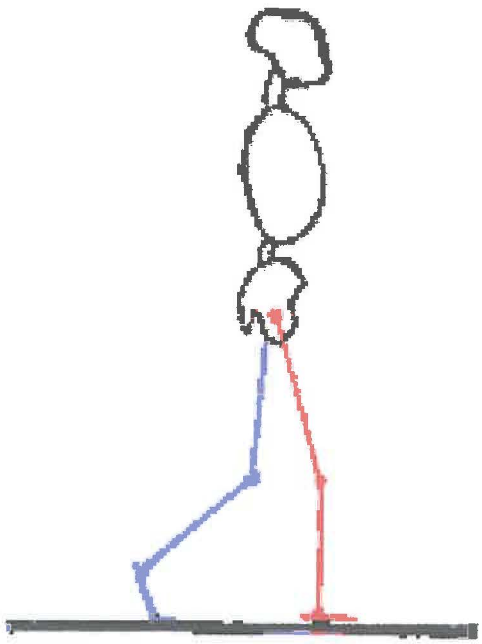

Figure 3. Stance phase, mid-stance 1. Obtained from: (Kaczmarska A. 2006)

4.2.3. Stance phase. (Loading response) Mid-stance:

- Single limb support interval
- Begins with the lifting of the blue foot and continues until body weight is aligned over the red (supporting) foot.
- The red leg advances over the red foot The blue leg is in its mid-swing phase
- Shoulder is in neutral
- Pelvis is in neutral rotation
- Hip is in neutral
- Knee is fully extended
- Ankle is relatively neutral
- Foot is pronated
- Toes are neutral

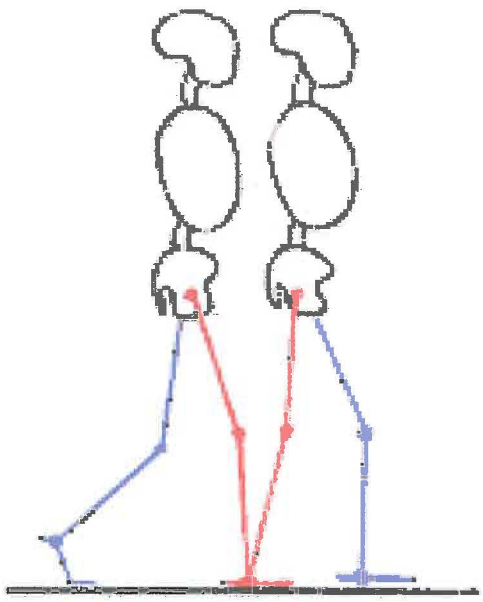

Figure 4. Stance phase, mid-stance 2. Obtained from: (Kaczmarska A. 2006)

4.2.4. Stance phase. Terminal stance:

- Begins when the red heel rises and continues until the heel of the blue foot hits the ground.
- Body weight progresses beyond the red foot
- Shoulder is slightly flexed
- Pelvis is rotated left
- Hip is extended and internally rotated
- Knee is fully extended
- Ankle is dorsiflexed
- Foot is slightly supinated
- Toes are neutral

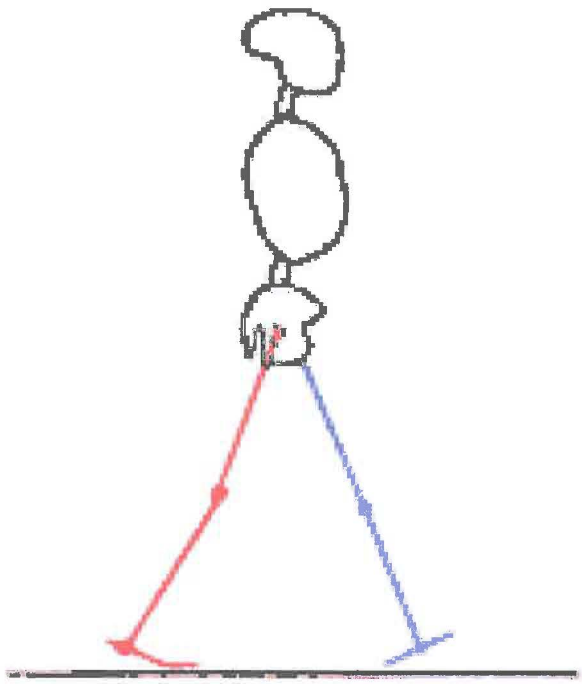

Figure 5. Stance phase, terminal stance. Obtained from: (Kaczmarska A. 2006)

4.2.5. Stance phase. Toe off:

- The second double stance interval in the gait cycle.
- Begins with the initial contact of the blue foot and ends with red toe-off.
- Transfer of body weight from ipsilateral to opposite limb takes place.
- Shoulder is flexed
- Pelvis is rotated right
- Hip is fully extended and internally rotated
- Knee is fully extended
- Ankle is plantarflexed
- Foot is fully supinated
- Toes are fully extended

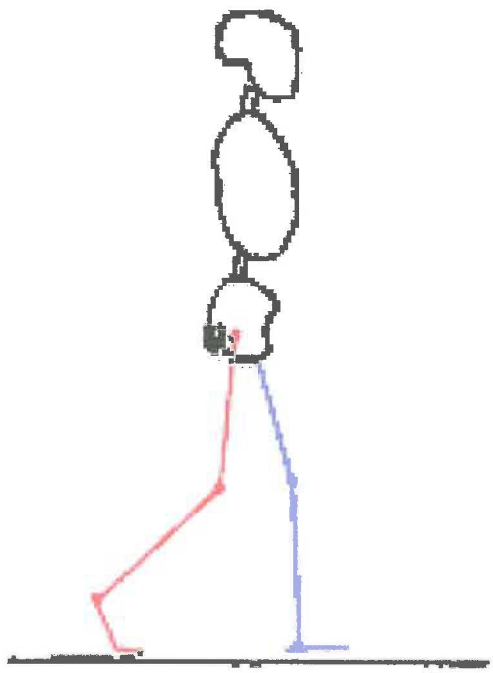

Figure 6. Stance phase, toe off. Obtained from: (Kaczmarska A. 2006)

4.2.6. Swing phase. Initial swing:

- Begins when the red foot is lifted from the floor and ends when the red swinging foot is opposite the blue stance foot.
- It is during this phase that a footdrop gait is most apparent.
- The blue leg is in mid-stance.
- Shoulder is flexed
- Spine is rotated left
- Pelvis is rotated right
- hip is slightly extended and internally rotated
- Knee is slightly flexed
- Ankle is fully plantarflexed
- Foot is supinated
- Toes are slightly flexed

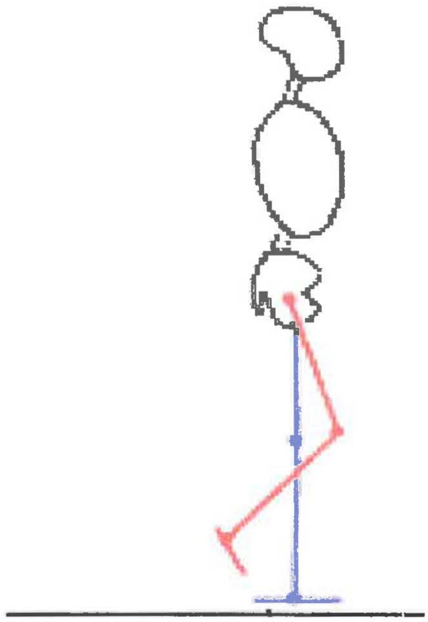

Figure 7. Swing phase, initial swing. Obtained from: (Kaczmarska A. 2006)

4.2.7. Swing phase. Mid-swing:

- Starts at the end of the initial swing and continues until the red swinging limb is in front of the body
- Advancement of the red leg
- The blue leg is in late mid-stance.
- Shoulder is neutral
- Spine is neutral
- Pelvis is neutral
- Hip is neutral
- Knee is flexed 60-90°
- Ankle is plantarflexed to neutral
- Foot is neutral
- Toes are slightly extended

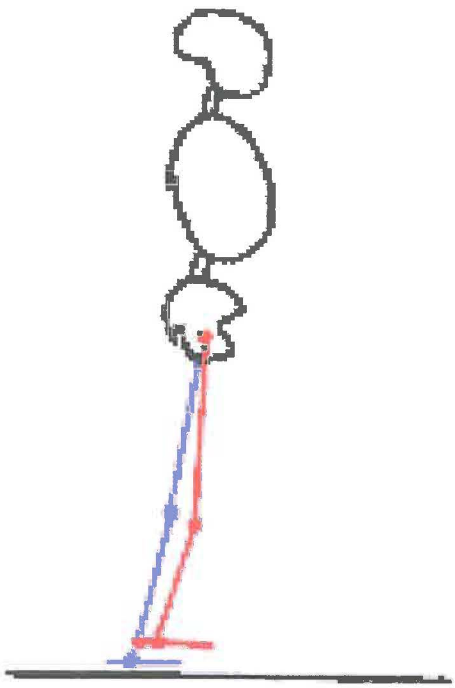

Figure 8. Swing phase, mid-swing. Obtained from: (Kaczmarska A. 2006)

4.2.8. Swing phase. Terminal swing:

- Begins at the end of midswing and ends when the foot touches the floor.
- Limb advancement is completed at the end of this phase.
- Shoulder is extended
- Spine is rotated right
- Pelvis is rotated left
- Hip is flexed and externally rotated
- Knee is fully extended
- Ankle is fully dorsiflexed
- Foot is neutral
- Toes are slightly extended

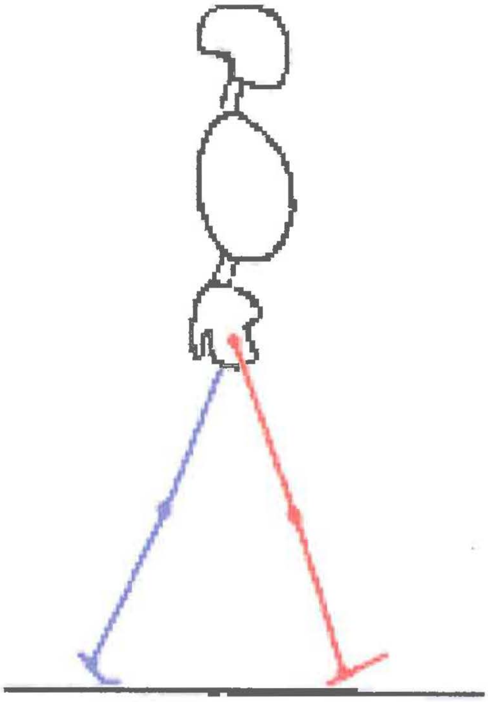

Figure 9. Swing phase, terminal swing. Obtained from: (Kaczmarska A. 2006)

4.3. Hip and torso movement during walking:

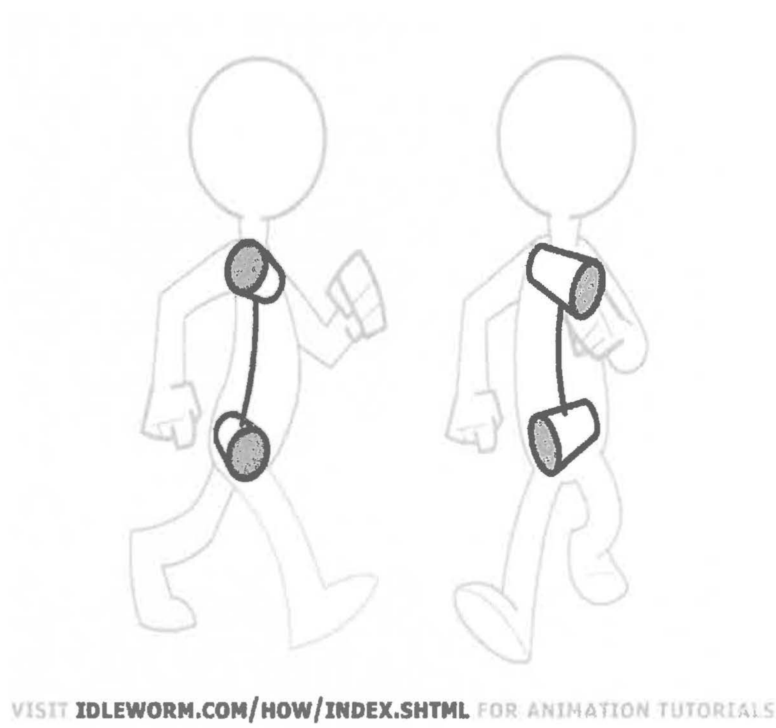

Figure 10. Hip and torso movement during walking. Obtained from: (http://www.idleworm.com/how/anm/02w/walk1.shtml)

In the above figure is illustrated the orientation of the shoulders and hips. Again, as one is thrust forward, the other is thrust back. As one tilts up, the other tilts down.

Another name for this is "Torque". It is a fundamental principle of good posing. It should be an element of almost every figure drawing that you do. Michelangelo always used torque in his sculptures, creating dynamic poses, even in ones that were standing still. One hip takes the weight, while the other passively provides the balance.

4.4. Gait – basic principles:

For initiation of gait the center of gravity (COG) shifts forward which leads to forward acceleration and as a result the step.

To stop the gait the center o gravity (COG) comes back into the foot basis which leads to deccelaration.

During walking, the COG is moving forward along inner side of the stance leg.

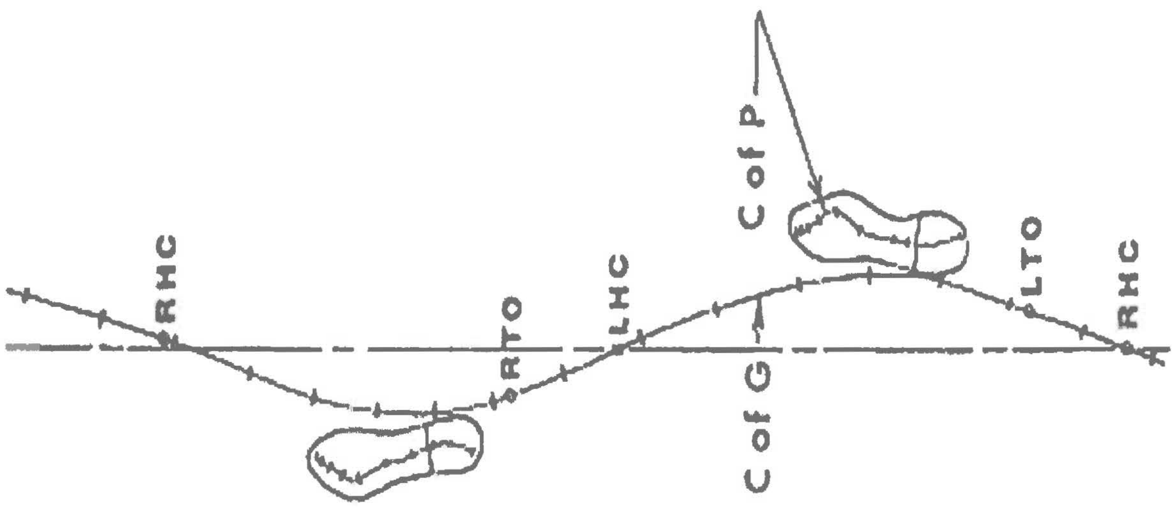

Figure 11. Shifting of center of gravity during gait. Obtained from: (Winter 1995)

5. Anatomic regions of the foot

The human foot is an amazingly complex piece of bioengineering. With 26 bones, 55 joints and a complex system of ligaments, muscles and tendons. The numerous joints between these bones allow the foot to be both a rigid lever and a shock absorber. At initial heel contact, the foot has to endure forces often in excess of 3 times body weight!

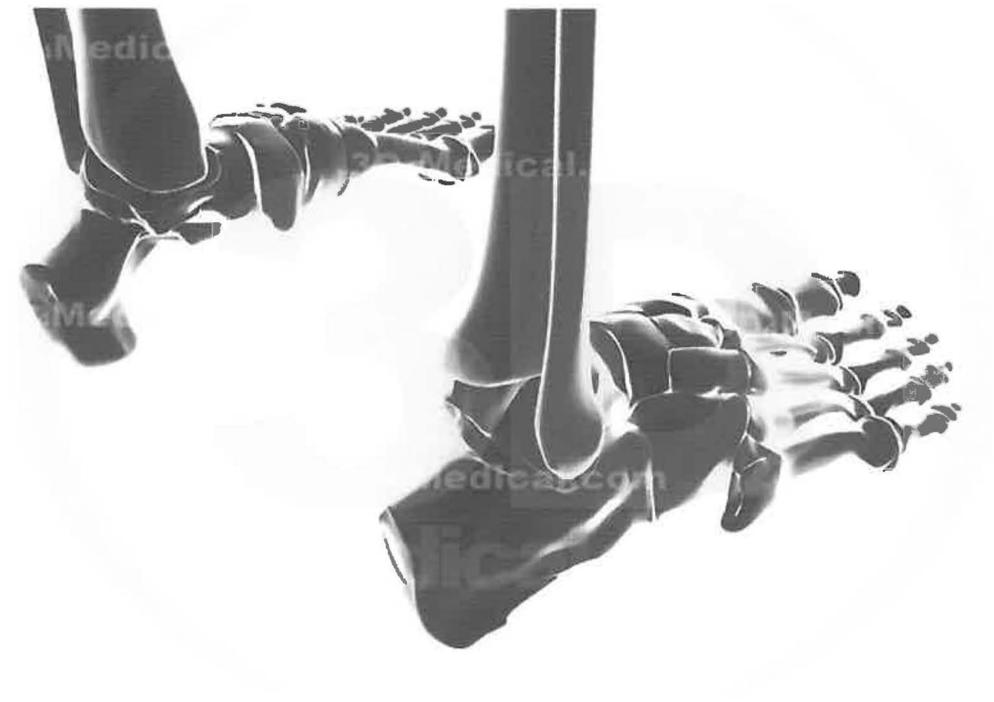

Figure 12. Bones of the foot. Obtained from: (http://images.3d4medical.com/The-bones-of-the-foot-73-image_RM5057.html)

The foot includes the area from the ankle through the toes. In some animals, including humans, the weight is supported on the entire surface of the foot. Such animals are known as plantigrade.

Like the hand, the human foot has five digits. However, it is less flexible and lacks an opposable digit (thumb) for grasping, as do the feet of most primates.

The human foot consists of 26 bones, connected by tough bands of ligaments. Seven rounded tarsal bones (the internal, middle, and external cuneiform bones, navicular, cuboid, talus, and calcaneus) lie below the ankle joint and form the instep.

Five metatarsal bones form the ball of the foot. There are 14 phalanges in the toes (two in the great toe and three in each of the others). The foot bones form two perpendicular arches that normally meet the ground only at the heel and ball of the foot. These arches are found only in humans. The use of the stride, a form of walking in which one leg falls behind the vertical axis of the backbone, is also a singular aspect of the human foot. The stride is thought to be an evolutionary advance from running, and is related to the unique structure of the human foot.

The forefoot includes the five metatarsal bones, and the phalanges (the toes). The first metatarsal bone bears the most weight and plays the most important role in propulsion. It is the shortest and thickest. It also provides attachment for several tendons. The second, third, and fourth metatarsal bones are the most stable of the metatarsals. They are well protected and have only minor tendon attachments and are not subjected to strong pulling forces.

Near the head of the first metatarsal, on the plantar surface of the foot, are two sesamoid bones (a small, oval-shaped bone which develops inside a tendon, where the tendon passes over a bony prominence) They are held in place by their tendons, and are also supported by ligaments.

The midfoot includes five of the seven tarsal bones (the navicular, cuboid, and three cuneiform). The distal row contains the three cuneiforms and the cuboid. The midfoot meets the forefoot at the five tarsometatarsal (TMT) joints. There are multiple joints within the midfoot itself. Proximally, the three cuneiforms articulate with the navicular bone.

The talus and the calcaneus make up the hindfoot. The calcaneus is the largest tarsal bone, and forms the heel. The talus rests on top of it, and forms the pivot of the ankle. (Kapandji 1974, Moore 1999, Netter 2001)

5.1. Foot and Toe Movement

Toe movements take place at the joints. These joints are capable of motion in two directions: plantar flexion or dorsiflexion. In addition, the joints permit abduction and adduction of the toes.

The foot as a whole (excluding the toes) has two movements: invertion and evertion. All the joints of the hindfoot and midfoot from the subtalar contribute to these movements, which are complex and consist of several components. In addition, foot movements ordinarily are combined with ankle movements. (Kapandji 1974, Moore 1999, Netter 2001)

5.2. The Arches of the foot

The foot has two important functions: weight bearing and propulsion. These functions require a high degree of stability. In addition, the foot must be flexible, so it can adapt to uneven surfaces. The multiple bones and joints of the foot give it flexibility, but these multiple bones must form an arch to support any weight.

The foot has three arches. The medial longitudinal arch is the highest and most important of the three arches. It is composed of the calcaneus, talus, navicular, cuneiforms, and the first three metatarsals. The lateral longitudinal arch is lower and flatter than the medial arch. It is composed of the calcaneus, cuboid, and the fourth and fifth metatarsals. The transverse arch is composed of the cuneiforms, the cuboid, and the five metatarsal bases.

The arches of the foot are maintained not only by the shapes of the bones as well as by ligaments. In addition, muscles and tendons play an important role in supporting the arches.

5.3. Muscles of the foot

The muscles of the foot are classified as either intrinsic or extrinsic. The intrinsic muscles are located within the foot and cause movement of the toes. These muslces are flexors (plantar flexors), extensors (dorsiflexors), abductors, and adductors of the toes. Several intrinsic muscles also help support the arches of the foot.

The extrinsic muscles are located outside the foot, in the lower leg. The powerful gastrocnemius muscle (calf) is among them. They have long tendons that cross the ankle, to attach on the bones of the foot and assist in movement. The talus, however, has no tendon attachments.

Figure 13. Muscles of the foot. Obtained from: (http://images.3d4medical.com/Muscular-foot-10-image_RM567.html)

Figure 14. Muscles of the foot. Obtained from: (http://www.3d4medical.com/)

The propulsive function of the foot depends on the arrangement of the bones into two longitudinal arches that act as shock absorbers; these arches flatten slightly when weight is put upon them and recoil when it is removed. In the standing position the weight of the body is mainly supported at the heel and the heads of the metatarsal bones, just behind the toes. On walking, the weight is first applied at the heel and then along the lateral border of the foot, medially across the metatarsal heads to the ball of the foot as the heel leaves the ground, and the big toe gives the final push-off. In running, the heel never touches the ground and the weight is applied only through the distal ends of the longitudinal arches, which recoil and reinforce the propulsive thrust delivered by flexing the medial toes. The arches of the foot are maintained by the shape of the interlocking bones, by muscle action, and by strong ligaments. If the ligaments become stretched and lax, part of the curvature will be lost, resulting in flat foot.

The foot is normally at right angles to the leg in the standing position. It can be drawn upwards (dorsiflexed) or lowered (plantar flexed) by movement at the ankle joint, and the sole of the foot may be turned inwards (inversion) or outwards (eversion); these movements involve the other bones of the foot swinging as a unit around the uppermost bone, the talus. (Kapandji 1974, Moore 1999, Netter 2001)

6. Plantar Pressure Distribution in Standing And Walking

Some variables of plantar pressure pattern have been investigated to establish a possible relationship between foot type and foot loading pattern. In one of a few studies investigating the effect of foot type on the plantar loading pattern during running, Sneyers et al.(1995) recorded plantar pressure distribution data in 24 (10 male and 14 female) athletes. The athletes were divided into pes planus, pes cavus and pes rectus groups. The height of the medial arch, the lower limb-heel alignment and the heel forefoot orientation assessed by static examination were used to divide participants into different groups. **The authors reported that the plantar heel load was directed towards the anterior part of the calcaneus in the pes planus group compared to the normal group. A relatively lower load under the midfoot for pes cavus, and a relatively higher load under the forefoot in pes cavus compared with pes planus were reported**, which are in agreement with the results of other studies. These findings generally support the proposed shift of load towards the forefoot in high arched supinated feet.

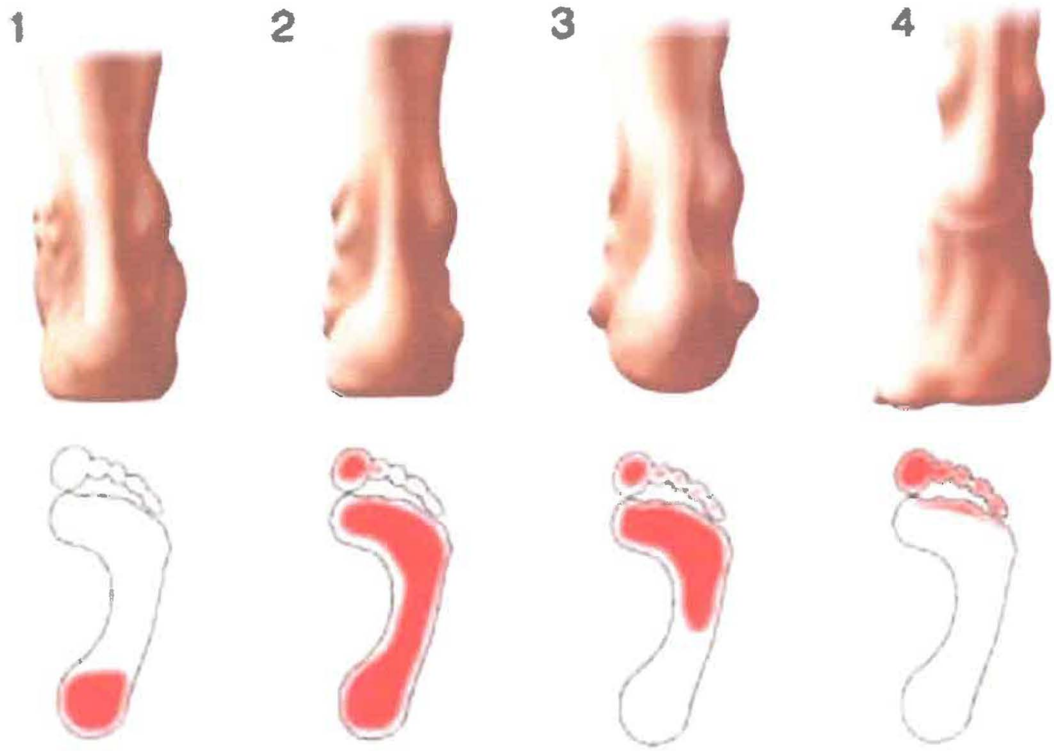

Figure 15. Normal pressure distribution during gait. Obtained from: (Winter, 1995)

In over-pronated feet, a medial shift of impulse and peak pressure under the heel has been reported, whereas the area of peak pressure under the metatarsal heads was found to be weakly related to foot type during walking. Other factors, including forefoot to rearfoot position, foot angle, step width and step speed, have been reported to be important.

Based on the results of their study, Walker and Fan (1998) argued foot type is a strong determinant of the pressure pattern. They justified the findings with the fact that the variables used to measure foot type and pressure were related to 2 different parts of stance phase of walking.

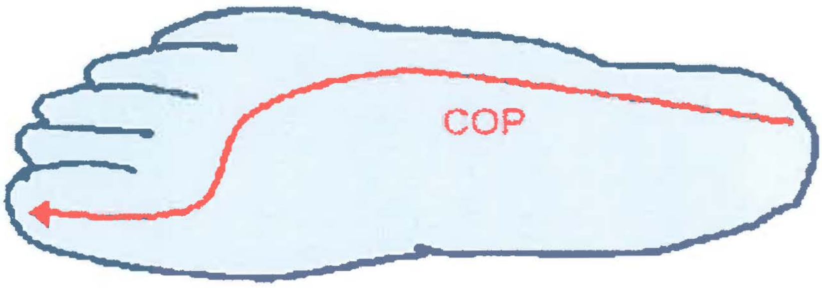

Figure 16. Normal trajectory of center of pressure (COP) during gait. Obtained from: (http://www.pt.ntu.edu.tw/hmchai/BM03/BMClinic/Walk.htm)

The role of shoes in altering the pattern of foot pressure and the site of peak pressure has been emphasised. Sneyers et al. (1995) **found statistically significant differences in most of the peak pressure ratios and impulses in all measured areas of all different foot types** in the barefoot compared with the shod condition. This was attributed to possible absorption of load by shoes.

In summary, it appears that different hindfoot to midfoot and/or midfoot to forefoot orientations and alignments used to classify feet in different research studies could inadvertently produce spurious pressure patterns. Different shoe types, surfaces and speed of walking and running may account for variation in findings. **Furthermore, relatively little is known about the effect of foot type on the plantar pressure distribution pattern, and the clinical importance of changes of plantar pressure pattern in different foot type needs to be more extensively investigated.**

7. Factors Which Effect The Foot Pressure During Walking

Physical activity is increasingly recognised as an important component of primary disease prevention. Overuse injuries are common sequelae of exercise and sporting activities in general, and of running in particular, frequently resulting in cessation of activity. It has been proposed that there is a link between foot shape, foot function and the occurrence of injury. As a means of treatment and prevention of further injury, orthoses and shoe inserts are widely prescribed in the belief that they can alter the pattern of lower extremity joints' alignment and movement. Although this is an assumption widely made in the treatment of many joint conditions, the manner through which this treatment could be effective is not clear.

The effects of foot type on the occurrence of lower limb injury during sporting activities and different aspects of biomechanics are reviewed, and the effects of applying orthoses on injury treatment and prevention and on various aspects of biomechanics of the lower limb joints are discussed.

Further research is required, firstly to establish the casual effect of foot type and function on the risk of lower extremity overuse injury, and secondly to document the specific effect of orthotic therapy on injury treatment and prevention. Specifically, more prospective studies are necessary to investigate the long term effect of orthotic intervention.

Biomechanical abnormality has been widely considered as an important aetiological factor predisposing running athletes to overuse injuries.

A difference in foot type, usually determined by the changes in the arch height of the foot, has been suggested to render athletes more prone to lower extremity overuse injuries. However, the mechanisms underlying the reported high incidence of running injuries associated with changes in arch height are not well established. The successful management of many sport-related injuries by the use of orthoses, reported in some clinical studies, has been deemed to lend further support to the belief that abnormal foot positioning during the contact phase of running could influence the function of the lower limb.

While this notion needs further scientific proof, the issue of effectiveness becomes more questionable by considering the following facts. **Despite apparent relief of symptoms from overuse injuries through the application of orthoses, a considerable percentage of athletes (up to 40% in some studies) so treated gain little or no benefit.** Indeed, increased symptoms and newly developed complaints have been reported during orthotic usage. This has been simply attributed to a poorly fitted and/or badly fabricated orthosis or poor diagnosis. Additionally, orthotic application has been shown to provide little relief of symptoms in patients with cavoid feet.

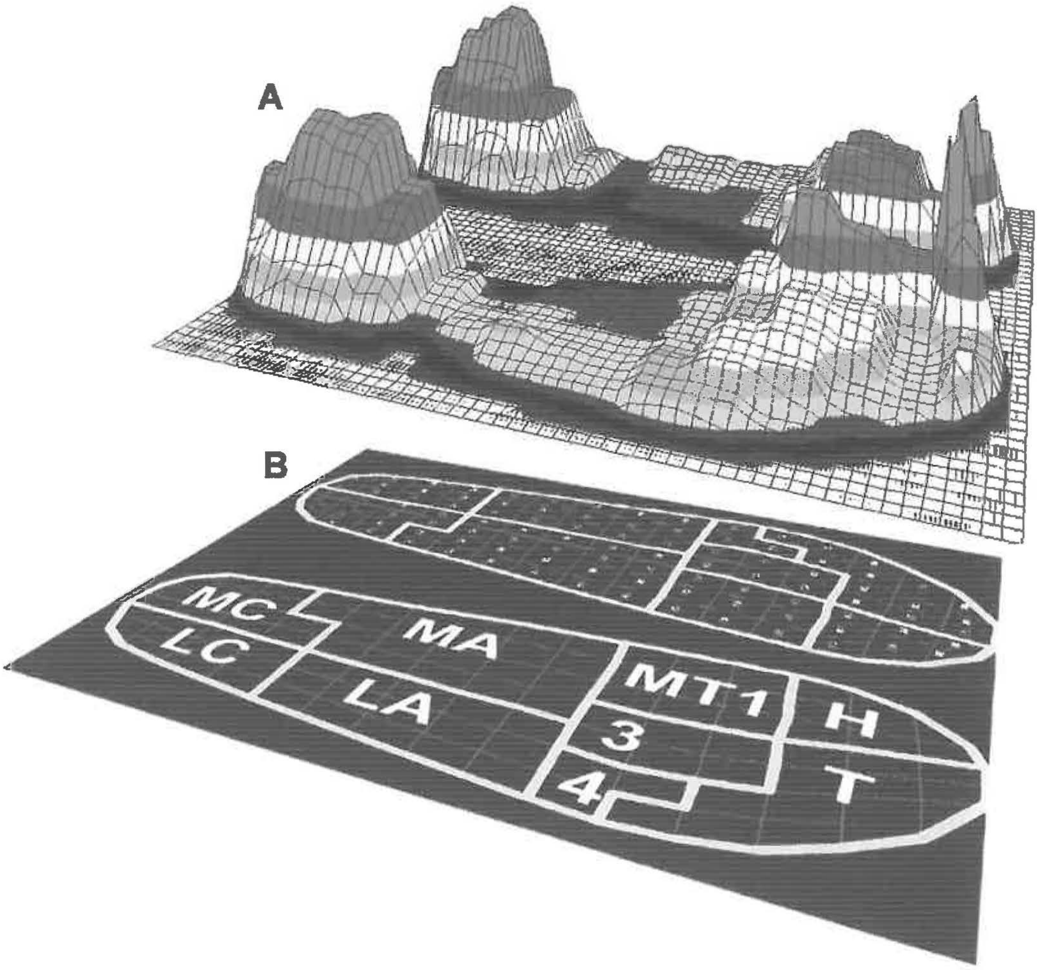

Figure 17. Foot pressure distribution. Obtained from: (http://www.biomedcentral.com/1471-2318/5/8)

Gaining an improved understanding of the interactions of foot type, injury occurrence and orthotic prescription necessitates finding evidence-based scientific answers to the following key questions:

- Is a specific foot type more prone to injury?
- Is a specific injury more common in the people with a specific foot type?
- How could such trends be explained?
- How is foot function related to foot structure?
- Which aspects of foot structure are related to foot function and gait?
- What is the relationship between static foot shape and dynamic foot and lower limb function?
- How valid are predictions of foot function based upon foot structure?
- Could the use of an externally applied device modify the movement and function of the foot?
- Could such an intervention improve the foot function, and thus prevent or treat overuse injuries?

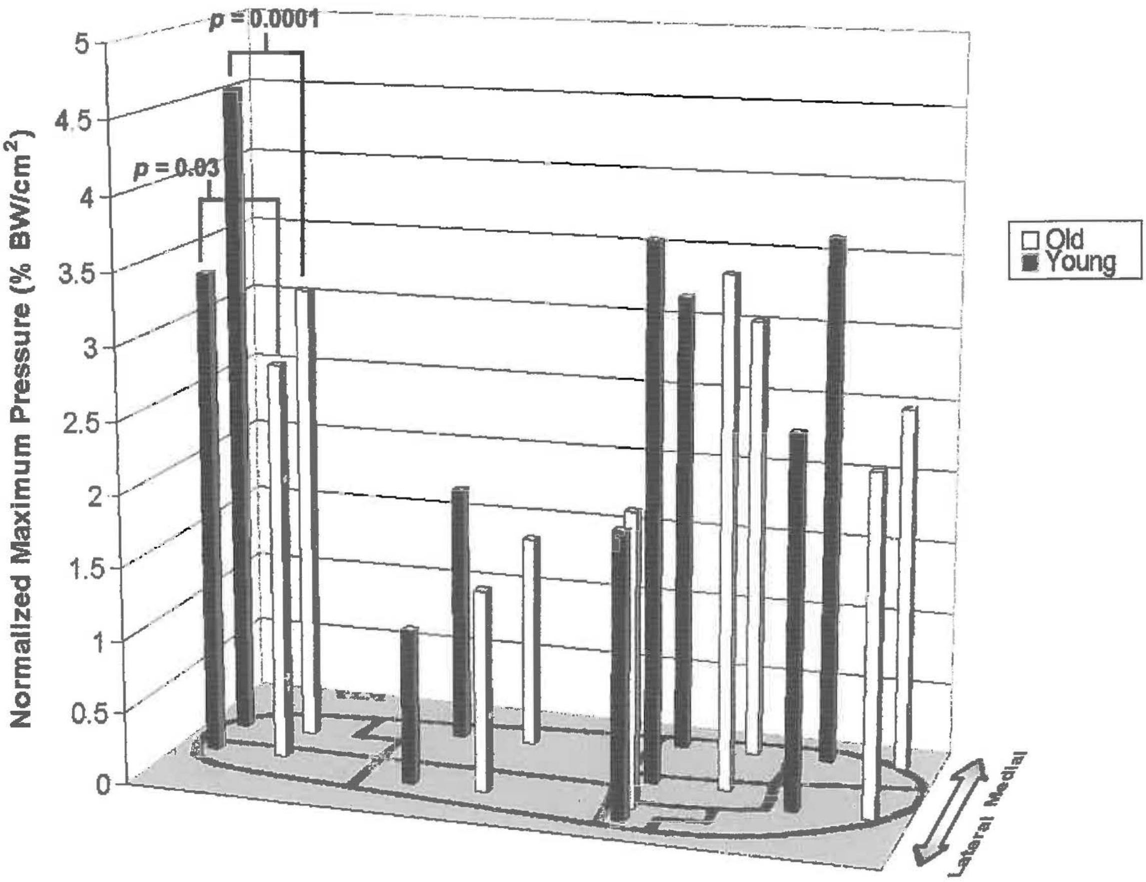

Figure 18. Pressure distribution by anatomical region between young ang old individuals. Obtained from: (http://www.biomedcentral.com/1471-2318/5/8)

Following a comprehensive review of the literature, clear answers to these questions remain elusive - there is little consensus on the issue of the relationship between foot type and injury and, furthermore, on the effectiveness of orthotic intervention in injury prevention and treatment. There is a lack of well-conducted research in this area. Specifically, the crucial role of potential confounders has widely been undermined or neglected in many studies.

7.1. Effects of Foot Type on the Occurrence of Sport Injuries

Higher risk of injury among physically active people has been reported for both low arched (flat or planus) and high arched (hollow or cavus) feet. Theories explaining such a relationship have been developed by considering the interrelated effects of foot type, subtalar mobility, rear-foot alignment and talocalcaneal/tibiocalcaneal movement relationship.

It is widely believed that a low arched foot tends to be more flexible and, thus, is subject to increased pronation (amount, timing and/or velocity) during the contact phase of walking and running. In contrast, a high arched foot is known to be more rigid and consequently exhibit increased supination. **Thus, a flat or cavoid foot may theoretically place the runner at a higher risk of injury. However, two successive prospective studies by Cowan et al. (1989, 1993) on US Army trainees provided no convincing evidence that low arched feet were more prone to injury; rather, it was suggested that low arched feet provide protection against lower-limb injury.** In both studies, a variety of measures of arch height were taken directly from photographs of feet while the volunteers were standing.

In a recent retrospective study, Wen et al.(1997) found lower extremity alignment measures, including arch index (AI), heel valgus (HV) and leg length discrepancy, were not major risk factors for running injuries. The participants were a cohort of relatively low mileage runners suffering injuries to the back, hip, knee, lower leg, ankle and foot. Participants were examined to measure alignment using specific criteria. Using the same method of examination and the same criteria, a subsequent prospective study by the same investigators appeared to confirm these findings. Overall, minor variations in lower limb alignment did not appear to be major risk factors for overuse injuries in these runners. Some minor associations were noted (e.g. higher varus and tubercle sulcus angle of the knee being associated with shin injuries) but additionally the results showed that higher AI and HV might protect against knee and foot injuries, respectively. **These studies, which may have had limited power, demonstrate the difficulties of investigating the multi-factorial nature of running injuries. Consideration of the combined effect of intrinsic and extrinsic factors is required. Rather than demonstrating relationships, it is necessary to be able to show a causal correlation.**

A high arched cavoid foot is often suggested to be associated with a higher incidence of stress fracture. It is proposed that a greater amount of energy is transferred to the lower limb bones through the relatively stiff foot, which causes an increased risk of femoral and tibial stress fractures. In contrast, Simkin et al.(1989) suggested the greater energy absorption low arched feet compared with high arched feet explains the high incidence of stress fractures reported in the metatarsal bones of patients with low arched feet.

Excessive pronation has been the aetiological variable most commonly linked to overuse injuries. However, the evidence supporting this contention may be injury specific. Viitasalo and Kvist, (1983) Gehlsen and Seger (1983) and Messier and Pittala (1988) **found a greater range of rearfoot movement during running in athletes from a variety of sports with different lower limb complaints**, including shin splints, plantar fasciitis and iliotibial band friction syndrome, than normal groups.

Higher velocity of pronation has been considered as another determinant factor of abnormal foot biomechanics. It has been proposed that abnormal subtalar pronation, associated with flat foot, results in an unstable foot at the time when a rigid lever is required at toe off, imposing a greater load on the body. Conversely, high arched foot configuration is supposed to cause hypomobility of the subtalar joint with a subsequent decrease in the ability to absorb the forces imposed on the foot.

However, maximal eversion motion has been reported to be independent of arch height. Nawoczenski et al. (1998) found similar magnitude of calcaneal eversion in both low rear foot and high rear foot groups, when 20 recreational runners were assigned to either low or high rear foot groups based on the lateral calcaneal inclination, lateral talometatarsal and anteroposterior talometatarsal angles measured on plain radiographs. **These findings taken together suggested that a functional relationship between arch height and injuries does not exist through the influence of arch height on hindfoot eversion**. However, such a relationship is supposed to exist through the influence of arch height on the amount of foot eversion that is transferred to internal rotation at the ankle joint complex.

More recently, kinematic analysis of the subtalar joint has been employed to explain the relationship between arch height and injury by introducing the concept of 'coupling behaviour' of the leg and rearfoot. Two important aspects of the subtalar joint have been described. Firstly, the subtalar joint attenuates the impact load of ground reaction force and, secondly, it has a unique role in the transfer of axial rotation of the leg to the pronation and supination of the foot during support phase of gait. Thus, supination and pronation of the foot produces rotation in the segments both proximal and distal to the subtalar joint.

The function of the foot and the movement patterns of the lower extremity are believed to be related to the orientation of the subtalar joint axis. An axis closer to the vertical plane supposedly results in a greater proportion of abduction and adduction of the foot, whereas an axis closer to the transverse plane would permit greater inversion and eversion. **The orientation of the subtalar joint axis influences not only the degree and direction of talocalcaneal joint motion, but also movements extrinsic to the joint.**

Nawoczenski et al. (1998) suggested the predominant rotation demonstrated by each foot group in their study to be determined by combined subtalar joint and talocrural joint axis orientation. This favours calcaneal inversion and eversion for the low rearfoot structure and tibial medial and lateral rotation for the high rearfoot structure. **They further proposed that it is not necessarily a lack of motion that accounted for the associated musculoskeletal problems of the high arched foot; rather, a greater proportion of tibial axial rotation may cause problems.**

In summary, despite significant progress in the understanding of the kinematics and kinetics of the foot and ankle complex, mechanisms causing **overuse injuries in the lower extremities are still poorly understood**. The effect of foot type on the occurrence of lower limb injuries has not been the subject of well controlled studies and few, if any, casual correlations have been demonstrated. Further studies are required to identify factors, determine causation and finally modify these factors with reexamination injury incidence.

7.2. Effects of Foot Type on the Ground Reaction Force

Kinetic parameters have been investigated in an attempt to evaluate the effect of arch height on the ground reaction force in running. **Nachbauer and Nigg (1992) found no effect of arch height on selected variables of the vertical ground reaction force**, including total impact force. The finding was explained by considering the timing of the events occurring in the early contact phase of a running gait. It has been shown that the point of force application lies in the rear one-third of the shoe at the moment of the maximal vertical impact peak. The line of action of the resultant force suggests the transmission of the force through the heel pad, calcaneus and talus into the lower leg. However, the rigidity of these hind foot bones, opposed to those of mid and forefoot, may be not related to the height of the arch or to any other structural property of the medial arch. **Thus, there may be no relationship between arch height and rigidity of the foot in the early support (stance) phase of running.**

Even if different foot types could impart variable loads on the body, it has been proposed that the neuromuscular control mechanisms of the runner could re-adjust the body reaction, thus minimising loading. These mechanisms act in such a way that the external impact force, regardless of the initial amount, is adjusted to a tolerable magnitude. Such an explanation has been employed to explain slight changes in the vertical impact force when running shoes with different midsole hardnesses were compared, emphasising the adjusting role of neuromuscular mechanisms.

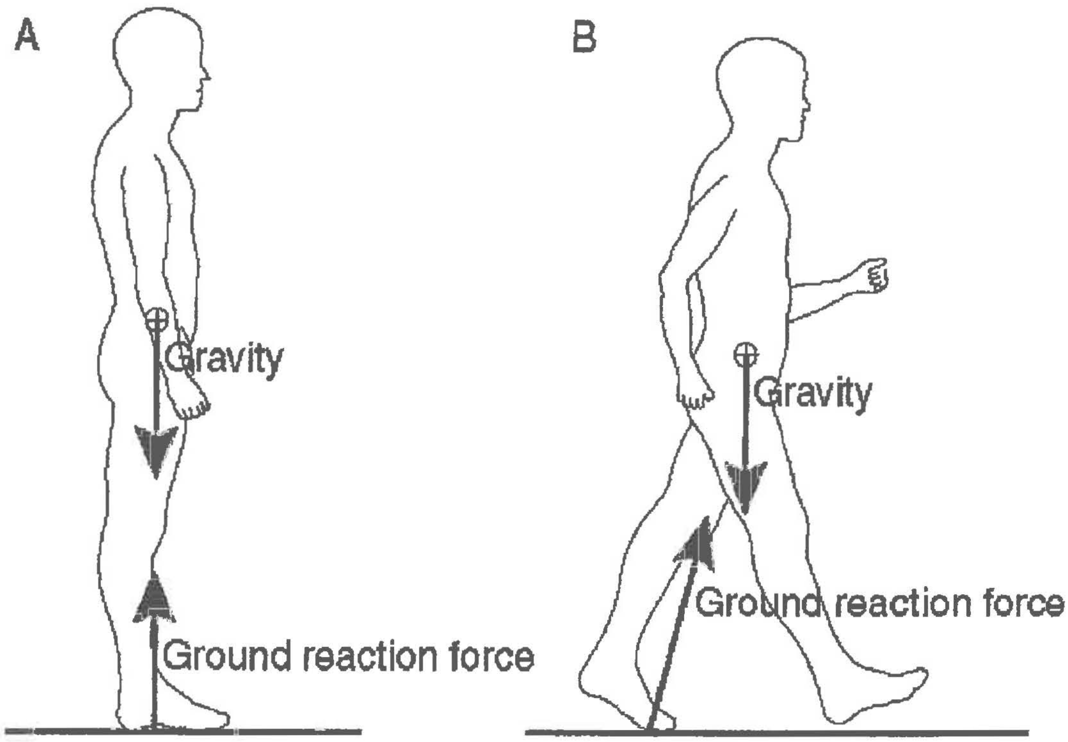

Figure 19. Ground reaction forces to which the foot is exposed during standing and walking. Obtained from: (Van Deursen 2001)

The direct articulation and coupling of the tibia/ fibula with talar motion relate rotation of the tibia to the inversion/eversion of the foot. 'Abnormal coupling' between the inversion/eversion movement of the foot and axial rotation of the leg is proposed as a contributing factor in lower extremity musculoskeletal injuries. **It has been suggested that certain knee and shin injuries are associated with an abnormal relative rotation of the tibia with respect to the femur.**Excessive tibial internal rotation is also believed to cause the foot to show abnormal movement patterns, specifically the amount, timing and velocity of pronation.

Nachbauer and Nigg (1992) compared selected ground reaction force variables in running for different conditions of arch height and arch flattening. Arch flattening was defined by the difference between the minimal vertical marker-floor distance in the midstance phase of running and the average vertical marker-floor distance while standing in the calibration frame. The arch height and arch flattening based on static and dynamic measurement of arch height were not found to be significantly related to each other. A significantly later initial medial force peak and lower anterior force peak in low arched and low flattening groups were reported, respectively. **Neither arch height nor arch flattening was thought to account for the observed variability in the ground reaction force.** Using a different method of classifying foot types by footprint parameters, **Hamill et al.(1989) reached the same conclusion.**

However, using a novel terminology, **Freychat et al. (1996) reported a relationship between rearfoot and forefoot orientation of the foot**, partially determined by the supinatory or pronatory position of the hindfoot, and ground reaction force parameters. The more the foot was 'open', the more it was placed closer to the direction of running. **They came to the conclusion that the specific spatial orientation of the rearfoot and forefoot can influence the 'open' and 'closed' foot behaviour**, making an open foot in flat foot configuration (laterally rotated, everted forefoot) more flexible. Whereas a medially rotated forefoot (closed foot) was associated with a rigid and inverted foot.

In a study of high and low arched feet, simultaneous measurement of amplitude and rate of impact loading at the ground and lumbar spine levels showed a lower magnitude of force in the high arched group in the spine, indicating a sound shock-absorbing capacity of high arched feet. There was no significant difference at the ground level between the 2 groups. **These findings are contradictory to those of Simkin et al.(1989) who showed an impaired shock absorbing capacity of high arched feet**, facilitating the transfer of shock wave to upper parts of the lower extremity. By assuming an association between initial pronation and an increased medial force component, individuals with high arched feet were found to have more unstable feet in the mediolateral plane at heel strike. A rapid internal rotation of the tibia immediately after touchdown was reported to be more likely to occur in high arched feet. The presence of this instability during impact loading, referred to as transient medial instability, rather than an intrinsically impaired shock-absorbing capacity was thought to explain the results.

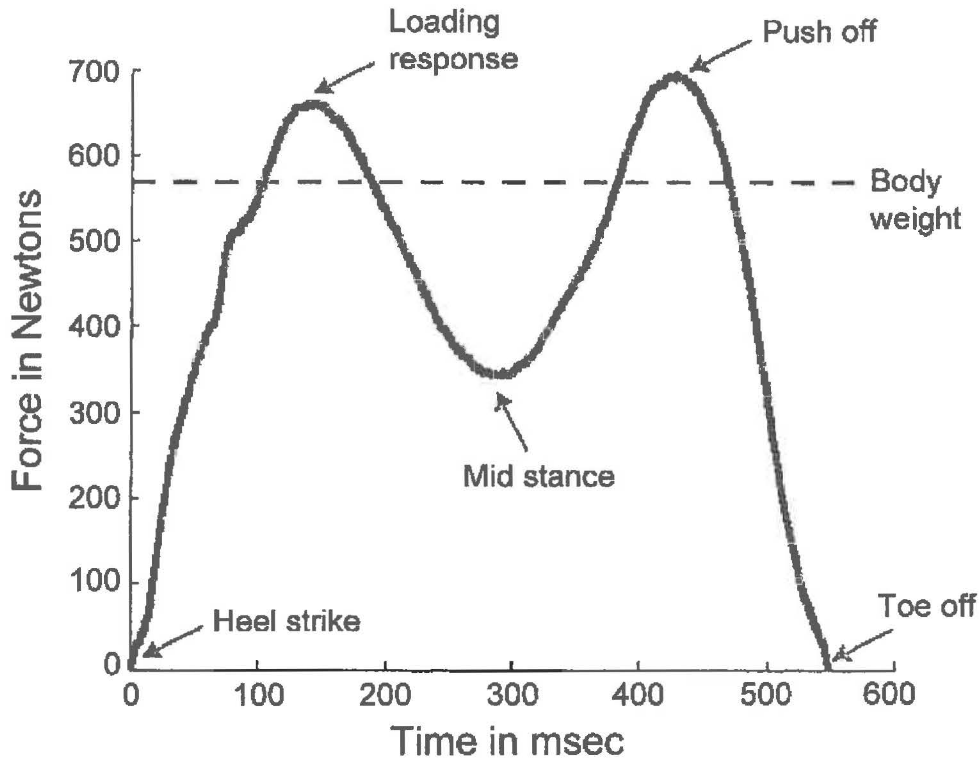

Figure 20. The vertical force component of the ground reaction forces during gait. Obtained from: (Van Deursen 2001)

In summary, the data presented above indicate little apparent difference in the magnitude of the vertical component of ground reaction force between different foot types, and that **any subtle differences that might exist are of dubious clinical relevance**. In the lower extremity, the shock wave resulting from the vertical component of ground reaction force is largely influenced by the position and orientation of joint axes, neuromuscular activity and muscle strength.

7.3. Effects of Foot Type on the Biomechanics of the Foot

To establish a better understanding of the functional relationship between arch height and injury, some investigators have attempted to evaluate the influence of arch height on kinematic variables of the lower extremity. **Clarke (1980) found a significant difference in the amount of pronation in an easy standing stance between high arched and low arched groups.** It is suggested that runners with flat feet spend a greater amount of time in pronation during the support phase of running. Repetitive loading caused by running with excessive subtalar joint motion is believed to render individuals with a flatter arch more susceptible to injury.

A greater range of motion in the subtalar joint has been measured in flat feet compared to high arched feet. **Subotnick(1985) suggested that high arched feet are inflexible, while flat feet tend to be hypermobile and susceptible to a large degree of pronation. However, in one study no difference was found in the rearfoot motion** between low and high arched feet during running. Nigg et al. (1993) examined the influence of arch height on axial rotation of the tibia. They proposed that such an influence might be expected because arch height could be an indicator of the structure of the tarsus, which acts as the link between the foot and the tibia. **Findings of this study suggested that arch height does not influence either maximal eversion movement or maximal leg rotation during running**. Rather, the transfer of foot eversion to internal leg rotation was found to increase significantly with increasing arch height.

By measuring foot placement angle (defined as the angle of orientation of the foot relative to the direction of travel) and AI, **Kernozek and Ricard (1990) reported that individuals with normal arches exhibited less total rearfoot movement than those with either flat or high arches**. Foot placement angle also had a negative relationship with total rearfoot motion. As foot placement increased, total rearfoot motion tended to decrease.

However, foot placement angle was found to be a better indicator of maximal pronation than arch type. As foot placement increased, maximal pronation decreased. **Arch height was found not to be a significant predictor of maximal pronation.**

It is believed that when the measurement of calcaneal inversion/eversion of the combined subtalar/talocrural joint is combined with the measurement of tibial internal/external rotation, a better insight to the kinematics of the subtalar joint is provided relative to single frontal plane analysis of each variable. **In summary, it appears that any investigation into the effect of foot**

type on variables of foot and ankle kinematics must take into account the interrelated function of subtalar, talocrural and knee joints.

7.4. Effects of Foot Orthoses on Overuse Injuries

Although orthoses may not provide a definitive cure for running overuse injuries, they are frequently prescribed for injury prevention. Logically, therefore, an evidence base to support the contention that orthoses are effective should exist. **However, little robust scientific evidence is available to support this notion.** Providing scientific evidence of the actual effect of orthoses on structure and function of the foot has been central to some studies. Although orthoses are frequently prescribed in the belief that they correct the biomechanical dysfunction of specific joints of the lower extremity, research studies on the effect of orthoses on rearfoot motion have not shown dramatic changes in this parameter.

To date, studies on the effect of orthoses on overuse injuries can be categorised into 2 key areas: the effects of foot orthoses on relieving symptoms of overuse injuries and effects on the biomechanical function of lower extremity joints.

7.5. Effects of Foot Orthoses on Relieving Symptoms of Overuse Injuries

Although the mechanisms by which orthoses are sometimes effective are not fully understood, a significant reduction in lower extremity symptoms has been reported.

In a retrospective study of the effectiveness of shoe inserts in long-distance runners, **Gross et al. (1991) found foot orthoses very effective in providing symptomatic relief** of lower extremity complaints. The complaints for which orthoses were prescribed included a broad range of hip, knee, foot and ankle problems. Unfortunately, the type of orthoses used was not clearly indicated and details regarding presumed diagnosis and the presenting indication for orthotic usage were reported by study participants. By finding that 90% of the runners continued to use the orthotic devices even after resolution of their symptoms, the authors concluded a high degree of overall satisfaction. Results of treatment were independent of diagnosis or the runner's level of participation. Orthotic shoe inserts were most effective in the treatment of symptoms arising from biomechanical abnormalities, such as excessive pronation or leg length discrepancy.

In general, a satisfactory level of symptom relief from use of orthoses has been reported in overuse injuries. After wearing orthoses for 3 months, 81% of 43 patients with painful heels treated with a customised rigid plastic foot orthoses were reported to gain a complete symptom relief.A functional foot orthosis was found to effectively reduce pain by 80% in patients with plantar fasciitis. Orthotic devices have also been reported to hasten the duration of returning to full functioning in injured runners.

Despite these positive findings, orthoses have been shown to provide little symptom relief in other athletes. Gross et al. (1991) reported that 24.5% of study participants made slight or no improvement, and 13.5% experienced increased severity of symptoms or developed new complaints during the period of orthotic usage. This was attributed to poorly fitted orthoses or poor diagnosis. Furthermore, orthotic application has been reported to have little success in relieving symptoms in patients with cavoid feet.

7.6. Effects of Foot Orthoses on Biomechanics of the Lower Extremity Joints

The effect of different foot orthoses has been investigated through kinematics, kinetics and pressure pattern of the foot.

7.7. Lower Limb Kinematics

In the area of research into the effects of orthotic prescription on foot biomechanics, the effect of an orthotic intervention on lower limb kinematics has drawn the most attention. The source of excessive motion of the rearfoot that results in the use of an orthosis has been an extensive area of study. Factors such as running shoes with soft midsoles and accommodative surfaces have been proposed to induce greater pronation than normal.

Orthoses have been reported to modify selected variables of lower limb kinematic behaviour during the stance phase of walking and running. Such interventions have been used to bring pronation in an injured foot closer to that of the normally aligned foot. **Nigg and Morlock (1987) reported a reduction in maximum pronation or calcaneal eversion by using an orthosis.** This finding was in agreement with those of previous studies.

Maximum pronation velocity, time to maximal pronation and total rearfoot motion have all been reported to be reduced by an orthosis.

McCulloch et al.(1993) examined the interactive effect of foot orthotics and 2 walking speeds on the angular changes at the rearfoot, ankle and knee, and temporal events during stance phase of walking. **They measured a significant reduction in the degree of pronation throughout stance phase of walking, as well as an increase in the duration of stance time** as measured from heel strike to heel rise when study participants wore functional orthoses. The orthotic application did not significantly reduce the velocity of pronation during the first 20% of stance.

In an attempt to examine the effects of semirigid foot orthoses on 3-dimensional lower limb kinematics, Nawoczenski et al.(1998) recorded the pattern of changes of the variables on 20 recreational runners presenting with distinct structural foot characteristics. The runners were classified into low or high rearfoot profile groups, based on radiograph measurements. **A significant orthotic effect was shown for rotations occurring from heel contact to peak tibial internal rotation, as well as in the coupling** relationship between tibial axial rotation and calcaneal inversion/eversion. **A similar mean reduction (2°) in tibial internal rotation was seen in both groups. Wearing an orthosis produced no significant change in the frontal plane rotations for either group.**

The authors concluded that the maximum effect of orthotics was related to the changes in tibial axial rotation and was only seen in the first 50% of stance.

Medially posted, custom-made soft orthoses were shown to change transverse and frontal plane movements of the foot and ankle during treadmill walking and running in a group of patients with patellofemoral pain syndrome. All patients exhibited forefoot varus >6° and/or calcaneal valgus >6°. **No differences were found in sagittal plane movements.** The **frontal and transverse planes rotation of the talocrural/subtalar joint were reduced** 1 to 3° with orthotic application. **The orthosis was reported to reduce knee motion in the frontal plane during the contact and midstance phases of walking.**

However, the motion was increased during the contact and midstance phases of running. **It was concluded that these results indicate orthotic intervention is effective in changing the pattern of transverse and frontal plane motion of the foot and knee.**

In comparison, rigid, medially posted orthoses were reported to significantly reduce kinetic and kinematic variables, including maximum calcaneal eversion angle, total rearfoot movement, maximum calcaneal angular velocity and maximum eversion angle, in a group of individuals with pronated calcaneus (a minimum of 5° calcaneal eversion) while standing. Providing a decreased pronation angle in standing and during the loading response phase of the gait was suggested as the main mechanism through which an orthosis could relieve clinical symptoms.

To evaluate the immediate effect of orthotic treatment for flexible flat foot, Leung et al.(1998) recorded changes of kinetic and 2-D kinematic variables in 8 individuals in shod and unshod conditions. A force platform and motion analysis system with 2 video cameras was used to collect data on kinetics and kinematics, respectively.

Relatively little change in the collected force data was found, whereas the **changes in displacement data with the modified University of California Biomechanics Laboratory (UCBL) shoe insert were significant.** The results found to indicate the effectiveness of the orthosis on aligning the orientation and movements of the subtalar, ankle and knee joints, thus reducing the degree and duration of abnormal pronation. It could potentially decrease strain on the plantar ligaments and reduce abnormal tibial rotation.

7.8. Ground Reaction Force

In the evaluation of the effects of orthoses on ground reaction force variables, most investigators have concentrated on the vertical component (impact force) of the ground reaction force. Changes in impact force with different ground surfaces, shoe materials and design have been widely investigated. **Nigg and Morlock (1987) reported that changes in shoe heel flare did not alter the magnitude of impact force peaks in a group of 14 male runners**. By comparing the effect of wearing shoes with different midsole hardness, De Wit et al.(1995) reported that harder material produced a smaller vertical impact force that occurred with a higher loading rate. A significantly larger and faster initial inversion was found when the volunteers ran in shoes with a hard midsole. Nigg et al. (1988) compared the changes of vertical force peak, time of occurrence of vertical force peak, and maximum vertical loading rate in a group of heel striker runners when wearing running shoes with conventional insoles to those when using shoes with 4 different viscoelastic insoles. **No difference was found in variables describing the vertical impact forces, and kinematic and kinetic variables of the lower extremities** were not influenced. In a study of ground surface materials with different impact absorbency, Dixon et al. demonstrated maintenance of similar peak impact forces for different surfaces. This finding was explained by individual kinematic adjustments to variable surfaces between individuals, suggesting that responses to different surfaces are individual.

Because of the fact that the technical process of capture and processing of shearing force (horizontal component of ground reaction force) **measurements has not been developed sufficiently, no information is available about the changes of shear with the use of different sorts of orthoses.**

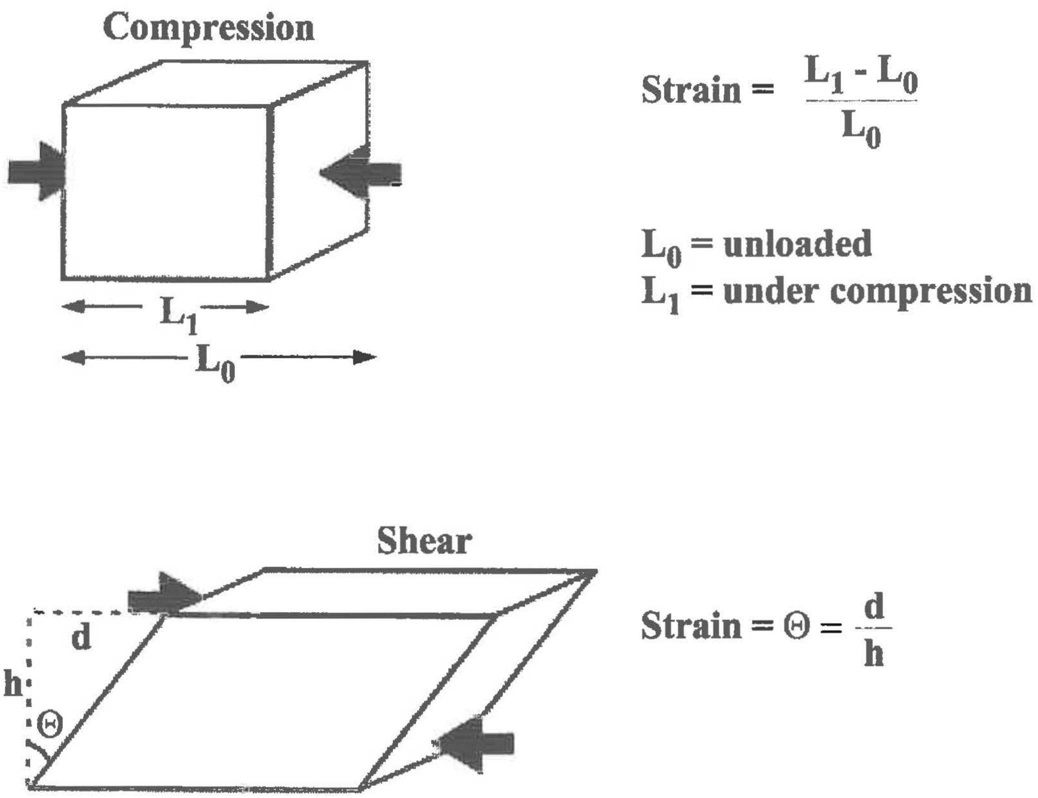

Figure 21. Calculation of strain. Obtained from: (Van Deursen 2001)

7.9. Foot Pressure Pattern

Little has been done in the area of research into the changes of foot loading pattern in orthotic management of overuse injuries. **A 30 to 40% reduction was reported in plantar pressure under the first metatarsal head and medial heel in patients with a custom-made orthosis.** The patients had a pronated foot type and underwent an in-shoe measurement to compare the total contact area under the foot between each patient with and without the orthotic device. The results were concluded to show the effects of the custom-made foot orthosis to increase total contact area (redistribute force) and, thus, to be able to reduce plantar pressures.

However, it has been now widely accepted that in order to provide a better understanding of the effect of orthoses on the foot loading pattern, it is essential to distinguish between the effect of an orthosis by itself and the interactive effect of shoe-orthosis on the foot function. **Thus, every orthotic intervention should be evaluated at 2 different levels: between the foot and the orthosis and between the orthosis and the shoe.**

Investigation into the absolute effect of orthoses is currently difficult because of the problem of simultaneously attaching measurement insole and the orthosis. Indeed, the examination of the effects in the 2 levels at once needs further technical development, which seems far more inaccessible at present.

8. Detection methods of foot shape and pressure distribution

8.1. Mechanical devices

Devices to record the force or stress beneath the foot have existed since 1882 when Beely had subjects step on plaster-filled sacks, theorizing that the magnitude of pressure was proportional to the depth of the impression. This method and others like it tended to record the shape of the foot and not necessarily the pedal forces.2-4 Another early method of recording stress was based on the deformation of pliable projections protruding from the underside of a mat upon which the subject walks or stands. This stress makes the projections collapse; the area of the mat in contact with the surface beneath increases, and produces a darkened area. The intensity of the inked area is proportional to the applied pressure. A pressure image is produced by an inked mat that

leaves a single peak pressure picture on the paper below the sole imprint.

The disadvantages of this method are twofold: one is the inability to provide any pressure versus time data, and the other is that the image reaches a maximum intensity after which no further increase in pressure can be detected.

The first mechanical device was Morton's kinetograph. The projections on this device consisted of longitudinal ridges that pressed an inked ribbon onto

a piece of paper and left a series of parallel lines that widened with increasing force.

Elftman used the principle of collapsing projections, but provided pressure-time data. His device consisted of a black rubber mat with pyramidal projections on the bottom that laid upon a glass plate. A white fluid filled the spaces between the pyramids and provided contrast when the pyramids spread. The image was recorded from below with a 16mm movie camera at 72 frames per second. This deformable projection principle is widely used today in the commercially available Harris mat. It has the advantages of being portable and providing better resolution than the previous devices, and is relatively inexpensive.

8.2. Capacitance transducers

A capacitance transducer consists of two conductive plates or elements separated by a flexible dielectric. As the pressure is applied to the device, the distance between plates decrease, the capacitance then increases, and its resistance to alternating current decreases.

Capacitance transducers may consist of a single layer of compressible material sandwiched between two conductive layers, or they may contain several capacitors in parallel by stacking several alternating layers of plates and dielectrics. **This type of device is inexpensive, stable, and produces fairly linear response, but tends to be thick, which makes it less adaptable for use in shoe transducers.**

In 1978 Nicol and Hennig were the first who developed a flexible matrix of capacitance transducers using a 48 x 24 cm foam-rubber mat with 16 conductive strips on either side. The strips were oriented orthogonally to form 256 transducers, 1 at each intersection of strips. The entire array could be scanned in about 5 ms.

8.3. Piezoelectric transducers

A piezoelectric transducer functions on the principle that certain crystalline structures are piezoelectrically active and function as a bundle of dipoles, with positive charges grouped at one side and negative charges at the other. When mechanical stress is applied to the material, separation of charge occurs proportional to the magnitude and orientation of the stress.

The advantages of this transducer are that smaller loads are produced under the foot and that output is linear and exhibits no hysteresis. Its disadvantages are that it is extremely sensitive to temperature changes. Also, the voltage decays with time, so the device is not suitable for static data collection.

There are many problems inherent in piezoelectric devices, problems which have discouraged clinical use of piezoelectric transducers.

Hennacy and Gunthe developed in 1978 the first piezoelectric transducer system. They used commercially available crystals (Vernitron PTZ-54) to build a piezoelectric pressure sensor that was easily calibrated, inexpensive, and capable of recording static and dynamic pressures.

8.4. Magneto resistor sensors

The magneto resistor uses a semiconductor theresistance of which varies with the strength of the magnetic field in which it is placed. The device was developed by Tappin to measure shear forces on the sole of the foot. The transducer is constructed using two stainless steel disks 16 mm in diameter. The upper disk is grooved and attached to the subject's foot.

The lower disk had a corresponding ridge which fit into the groove of the upper disk and allowed sliding translation between the two disks along one axis only. A magneto resistor is mounted flush with the floor of the groove, and a magnet is attached to the ridge. When assembled, the magnet and resistor would slide relative to each other. The disks are held together with silicone rubber, which allow translation of the disks relative to each other and provide a recentering force. The electrical signal produced is proportional to the movement of the magnet, which is in turn proportional to the applied shear force.

8.5. Foot Imprinter

The set includes the imprint mat, paper, ink, and roller. This kind of method produces weight-bearing imprint of the foot, thereby measuring pressure disbursement and arch shape.

Figure 22. Foot Imprinter. Obtained from:(http://www.mmarmedical.com/images/Apex-Imprint.gif)

This method is a simple tool for the measurement and modification of plantar pressure points. Gives clear picture of the plantar pressure points. It is an excellent educational tool to motivate the patient for the better progression of the therapy.

The advantage of this method is tha it is simple, yet effective, and with minimal cost.

The foot imprinter displays weight distribution on the plantar surface of the foot. It can easily be used in dynamic gait analysis, static weight bearing and non-weight bearing positions.

The examiner applies ink on the underside of the mat and places a blunk paper underneath the mat. Then the patient steps on the mat and his/her footprint is printed on the paper. No ink comes in contact with patient or operator.

8.6. Podoscope

Podoscope is a diagnostic device which is used for evaluation of foot problems. The evaluation is direct. Measurement of foot size, heel position, toes position, position, shape of the arches and overloaded points under the foot are easily diagnosed. Integrated lamp.

Advantage of this method is tha it is simple, yet effective, and with minimal cost.

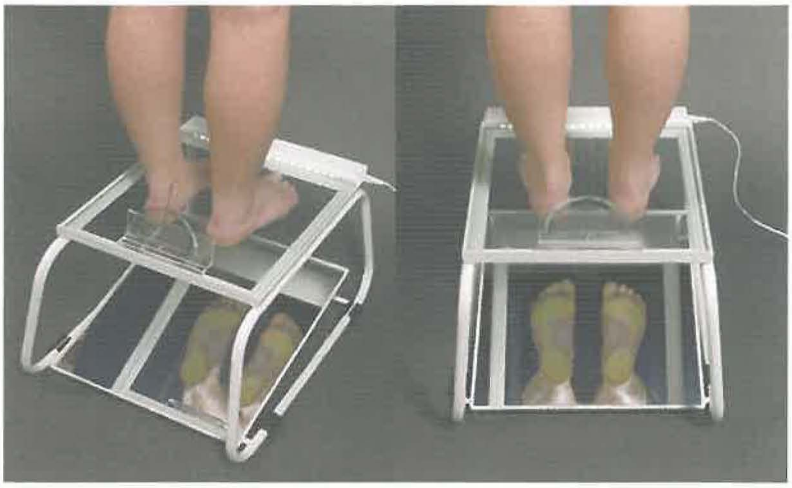

Figure 23. Podoscope. Obtained from: (http://www.ingcorporation.cz/img/news/P4285808.jpg)

The podoscope is a device designed to assess the interaction of the foot and supporting surface. A patient stands on the transparent glass plate of the podoscope's and the image of his feet shows through mirror to the person who is doing the measurements.

8.7. Foot scanning methods

The foot scanning method record a 2-D scan of the bottom outline shape of the foot.

Figure 24. Foot scanning. Obtained from: (http://www.shoemaster.co.uk/solutions/ecofoot.jpg)

Laser scanning technology has vastly improved the process of fitting foot orthotics. **Advantages of scanning far outweigh traditional casting methods. Foot scanning is more accurate than casting, leaving little room for error or interpretation. Foot scanning is faster and cleaner.** A complete scan takes only few minutes. Furthermore, it is easy for the foot care professional to learn how to scan feet.

8.8. In-shoe dynamic pressure measuring system

The foot platform analysis system enables professionals to perform **dynamic pressure profiling** in order to evaluate shoe-to-ground interaction related to the diabetic foot for example, pronation, foot arch and weight-bearing capabilities and assess impact effects in bipedal locomotion activities of both feet, either exclusively, or in relation to each other. The platform detects body motion (foot-knee-hip) to effectively profile any abnormalities. In addition, densely packed sensors in the platform analysis system offer the user high resolution images and a modular architecture.

TACTILUS® INSOLE SYSTEM HARDWARE
(COURTESY OF SENSOR PRODUCTS LLC 2005)

Figure 25. In-shoe pressure measuring system. Obtained from: (http://ww1.prweb.com/prfiles/2005/02/24/212471/FootInsolewithcaption.jpg)

The insole systems are advantageous in their basic design by assessing foot-to-shoe interaction. The foot insole is comprised a thin and highly durable substrate material and ranges in size. The insole sensor collects precise data for determining pedal pressure points and assessing athletic plantar implants in activities ranging from standing and walking to running, jumping, skiing and skating. This kind of system works at speeds up to 500 Hz.

Both the foot platform and insole analysis systems possess robust sensors which can endure thousands of uses with consistent repeatability, and are highly resistant to electromagnetic noise, temperature, and humidity fluctuations. With special software provides isobar and region-of-interest viewing, graphical displays of data in bar charts, line scans and histograms, statistical analysis of average/minimum/maximum pressures, total force over any selected area, analysis view of all nine major foot zones, pressure versus time and more.

8.9. Pressure transducers in shoe sole

These transducers record the heel and toe strike activity as the subject walks. This system uses two force sensitive resistors (FSR). Typically one FSR is attached to the sole of a shoe at the heel and the other FSR is attached at the toes. The FSRs indicate the precise pressure placed on the heel and on the toe as the subject walks. This system comes equipped with a 7.6-meter cable and is designed for direct connection to the receiver module.

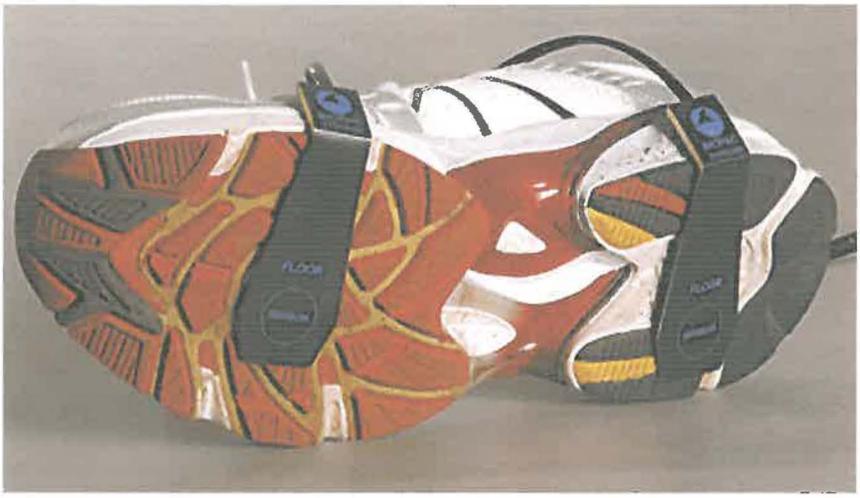

Figure 26. Pressure transducers in shoe sole. Obtained from: (http://www.biopac.com/ProductImages/ss28.gif)

8.10. Plantar pressure platform

The distribution of barefoot plantar pressure is measured using the pressure platform in the figure below. This pressure plate, placed on the walkway level with the floor, contains an array of 6,080 high quality capacitance sensors. Each sensor has a surface area of 0.25 cm^2 and can record pressure from 0 to 127 N/cm^2 during posture or comfortable cadence locomotion. The data is collected at approximately 50 samples per second and analyzed on a Pentium microcomputer.

Figure 27. Plantar pressure platform. Obtained from: (http://www.novel.de/productinfo/systems-emed.htm)

Using the special software, peak pressure, pressure time integral, and maximum force are quantified for different regions of the foot. In addition, a custom-developed program is used to calculate certain parameters of interest, including the Center of Pressure Excursion Index (CPEI), peak pressure, foot angle, and the temporal sequence of loading for the three phases of stance.

The primary value of a platform-based pressure distribution analysis is to objectively document dynamic barefoot function (e.g. excessive pronation) and aberrant pressure distribution during gait.

The measurement method is based on calibrated capacitive sensors. These systems are able to record dynamic as well as static measurements. The dynamic measurement is the most important since it determines loading during the actual roll-over processes, quantified parameters such as length and width changes of the foot, the Varus or Valgus position, the contact area of the foot, the function of the toes, joints and ligaments as well as other parameters.

The sensitivity of the sensors in some platforms is adjustable and the sensor can be calibrated to convert output into pressure units, such as PSI or mmHG.

These platforms vary in the dimensions. The small ones evaluate only one step each time but there also others that have the ability tou evaluaty a whole gait cycle which is a very important because we get additional data about the examining person.

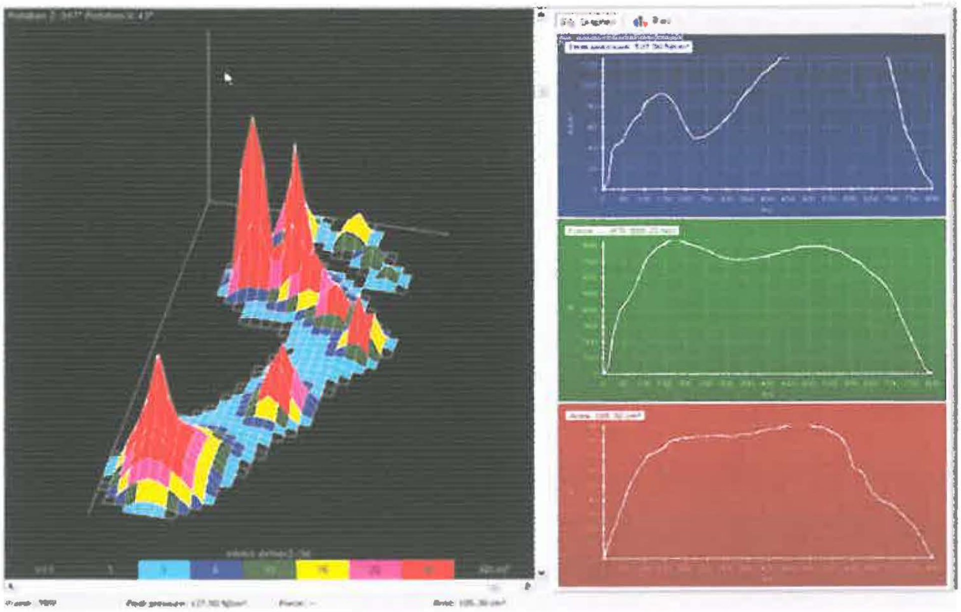

Figure 28. Screen shot of EMED data showing the vertical force component of the ground reaction forces. Obtained from: (http://podiatry.temple.edu/gaitlab/facilities/images/emed_screen.gif)

8.11. Pedobarograph

The study of Biomechanics comprises the processes involved in the movement of living beings. The use of force plates supply precise information on the magnitude and behaviour of the forces involved in these processes. It is designed to measure the three orthogonal components of the resultant force acting on the platform, and the three components of the generated moment in the same orthogonal co-ordinate system.

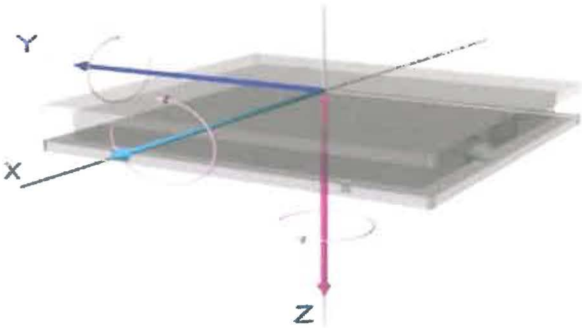

Figure 29. Co-ordinates of force plate measurements. Obtained from: (http://www.soe.uoguelph.ca/webfiles/cwse/Images/People/girls-on-pedo.gif)

As a pedobarograph the instrument is used as a gait analysis tool that measures the pressure distribution on the bottom of the foot through all stages of the gait cycle. The optical pedobarograph in the Biomedical Engineering lab uses digital video capture technology to record the pressure variations on the sole of the foot. The subject walks across the force plate fitted with an illuminated glass plate. As the foot hits the device, the glass surface deflects due to the force, causing the horizontal light beams to reflect downwards and be read by the video camera. The amount of light reflected is proportional to the pressure caused by the foot striking the plate.

Figure 30. Pedobarograph. Obtained from: (http://www.amtiweb.com/images/OR6-WPgrey.JPG)

8.12. Optic pedobarograph

A video pedobarograph system for providing a real time, qualitative display of dynamic relative pressure measurements includes a plurality of force sensors, a substantially rigid support structure and video pedobarograph electronics. The force sensors generate dynamic relative pressure signals and are positioned within a force sensor matrix structure. The substantially rigid support structure includes a substantially planer surface to which the sensor matrix structure is fixedly secured. The video pedobarograph electronics include a video sync stripper and control logic. The video sync stripper strips a video sync signal from a composite video signal received by the video pedobarograph electronics. The control logic maps the dynamic relative pressure signals to the composite video in response to the video sync signal to generate a mapped composite video signal providing a qualitative display of the dynamic relative pressure signals within a predetermined portion of an overall video image generated from the mapped composite video signal.

Figure 31. Optic pedobarograph. Obtained from: (http://www.amtiweb.com/images/OR6-GTgrey.GIF)

The optical pedobarograph has proven to be very helpful in studying foot pressure abnormalities in a variety of clinical conditions and especially in diabetes mellitus and rheumatoid arthritis. Studies using this device have provided a very good insight into the etiopathogenesis and natural history of foot disorders. It has also allowed the conduction of intervention trials which assess the efficacy of new treatment. The main advantages of the pedobarograph include accuracy, reliability and high spatial resolution. Its drawbacks are its size and that it can only measure pressures between the foot-floor interface.

Other specific uses include, stability analysis, neurological analysis, prosthetics fitting, athletic performance, shoe design, tire testing, force, power and work studies.

8.13. 3-D foot scanner

This kind of method used for foot shape remodelling might be also portable. It has laser projectors and video cameras mounted on the system. When the scanner is operated, laser lines make a cross section of the surface of the foot. The entire foot shape, including the sole, can be measured by taking pictures from under the foot as the subject stands on a glass surface. As the laser lines scan the foot, the projected cross section images will be recorded by the cameras, and the 3-D foot shape can be measured.

Figure 32. 3-D foot scanner. Obtained from: (http://www.iwl.jp/main/infoot_std.html)

By itself, raw measurement data is just a collection of points. Human anatomical information needs to be added for this collection to be handled as human data. For this reason, the human body's anatomical landmarks must be measured at the same time. The device extracts the points marked by a special marker as anatomical landmarks. Furthermore, it cross-checks against the foot shape database and automatically labels what kind of anatomical landmarks those points are.

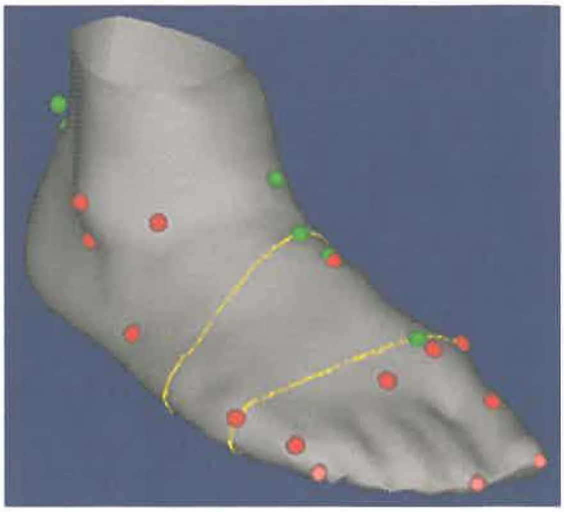

Figure 33. Surface model of the foot. Obtained from: (http://www.iwl.jp/main/infoot_std.html)

Based on the location data of these landmarks, some foot measurements will be automatically calculated. There is a significant difference between the data obtained by this measurement system and manual measurement by an expert anthropometrist. However, the difference in the results is smaller than 2.0[mm].

Also, by using a special software, a homologous shape model of the foot is automatically calculated based on anatomical landmarks. As a result, it will be easier to process shape information in addition to measurements. The result can be used for comparison of individual differences, designing of shoe lasts, and statistical analysis.

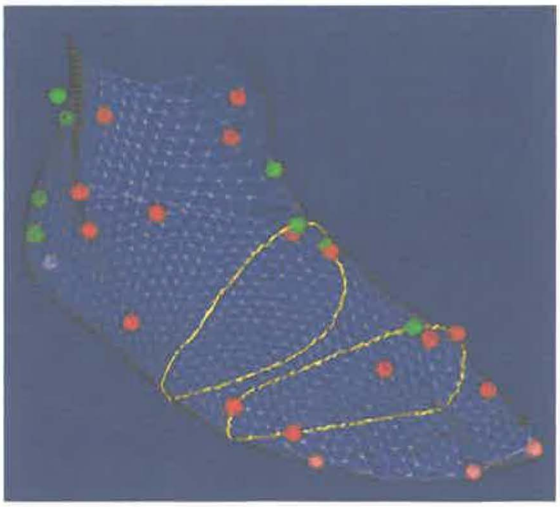

Figure 34. Polygon mesh of the foot. Obtained from: (http://www.iwl.jp/main/infoot_std.html)

8.14. 3-D Footmodeller

With the 3-D Footmodeller feet, plaster casts or lasts can be digitised in 3-D. Because the elastic foil encloses the object perfect, it makes a 100% copy of the object, which is digitised. Feet of patients can be scanned standing up or sitting down. The scanner scans the feet up to 3,5cm with an accuracy of 0,5mm. Within 10 seconds the measurement is made and because of it's shape and weight, it is very mobile. The scan data is compatible with the Insole-King Cad software, therefore it can be adjusted for designing insoles.

Figure 35. 3-D Footmodeller. Obtained from: (http://www.insole-king.com/insole-king/02-IK-Scan.htm)

8.15. Stereophotogrammetry

Stereophotogrammetry is the science of dimensional analysis of photographs using stereoscopic methods and equipment. It is a standard procedure used by land surveyors in the preparation of topographic maps from aerial photographs. The technique, which is noninvasive, has been used in a number of applications in medicine, orthopaedics, and oral surgery.

If aplied on the surface of the foot it gives us a very nice topographic map of the foot. It can be used to evaluate especialy the arches of the foot.

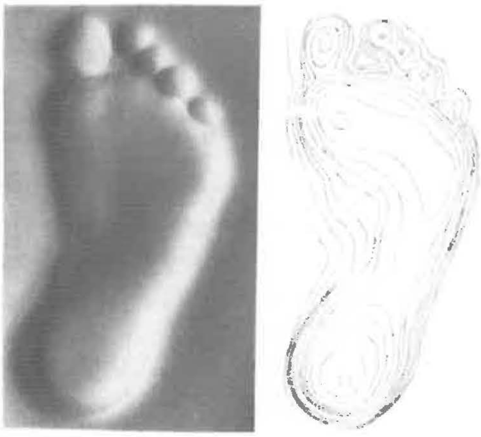

Figure 36. Stereophotogrammetry of the foot. Obtained from:
(http://www.pubmedcentral.nih.gov/picrender.fcgi?artid=1000663&blobtype=pdf)

8.16. 3-D Finite Element Modeling of the Human Foot

In order to provide a supplement to the experimental inadequacy, many researchers had turned to the computational methods in search of more clinical information. Computational modeling, such as the finite element (FE) method has been used increasingly in many biomechanical investigations with great success due to its capability of modeling structures with irregular geometry and complex material properties, and the ease of simulating complicated boundary and loading conditions in both static and dynamic analyses. The FE method can be an adjunct to experimental approach to predict the load distribution between the foot and different supports, which offer additional information such as the internal stress and strain of the ankle-foot complex.

The FE analyses could allow efficient parametric evaluations for the outcomes of the shape modifications and other design parameters of footwear without the prerequisite of fabricated footwear and replicating patient trials.

Existing FE models of the foot or footwear in the literature (Bandak, 2001; Barani, 2005; Camacho, 2002; Chen, 2003; Chu, 1995; Gefen, 2000; Giddings, 2000; Goske, 2005; Jacob, 1999; Lemmon, 1997; Lewis, 2003; Nakamura, 1981; Shiang, 1997; Syngellakis, 2000; Verdejo, 2004) were developed under certain simplifications and assumptions such as a simplified or partial foot shape, assumptions of linear material properties, infinitesimal deformation and linear boundary conditions without considering friction and slip. Although several 3D foot models were developed recently to study the biomechanical behaviour of the human foot and ankle, a geometrically detailed and material realistic 3D FE model of the human foot and ankle specialized for footwear or orthotic design has not been reported.

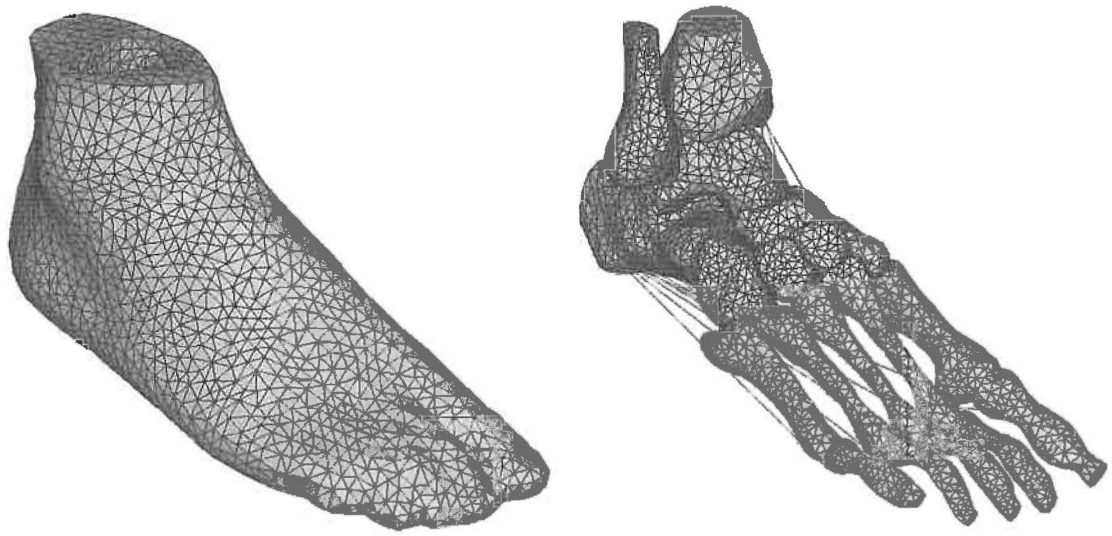

Figure 37. Surface model and finite element meshes of the encapsulated soft tissue and bony structures of the foot. Obtained from: (Camacho 2002)

8.17. 3-D Motion capture system

Gait analysis is the major application of motion capture in clinical medicine. Motion tracking or motion capture started as a photogrametric analysis tool in biomechanics research in the 1970s and 1980s, and expanded into education, training, sports and recently computer animation for cinema and video games as the technology matured. The examined person wears markers near each joint to identify the motion by the positions or angles between the markers. Acoustic, inertial, led, magnetic or reflective markers, or combinations of any of these, are tracked, optimally at least two times the rate of the desired motion, to submillimeter positions. The motion capture computer software records the positions, angles, velocities, accelerations and impulses, providing an accurate digital representation of the motion.

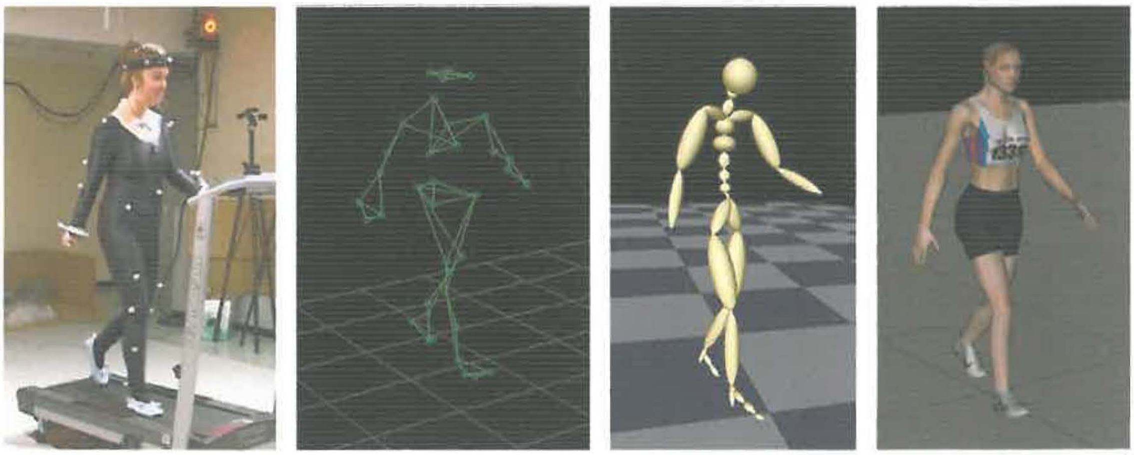

Figure 38. Motion data acquisition. Obtained from:
(http://vrlab.epfl.ch/research/images_research/LO_lorna_mocap.jpg)

Passive optical system use markers coated with a Retroreflective material to reflect light back that is generated near the cameras lens. The camera's threshold can be adjusted so only the bright reflective markers will be sampled, ignoring skin and fabric.

In biomechanics, sports and training, real time data can provide the necessary information to diagnose problems or suggest ways to improve performance, requiring motion capture technology to capture motions up to 140 miles per hour for a golf swing.

Optical systems utilize data captured from image sensors to triangulate the 3D position of a subject between one or more cameras calibrated to provide overlapping projections. Data acquisition is traditionally implemented using special markers attachedthe examined person.

These systems produce data with 3 degrees of freedom for each marker, and rotational information must be inferred from the relative orientation of three or more markers, for instance shoulder, elbow and wrist markers providing the angle of the elbow.

The centroid of the marker is estimated as a position within the 2 dimensional image that is captured. The grayscale value of each pixel can be used to provide sub-pixel accuracy.

An object with markers attached at known positions is used to calibrate the cameras and obtain their positions and the lens distortion of each camera is measured. Providing two calibrated cameras see a marker, a 3-dimensional fix can be obtained. Typically a system will consist of around 6 to 24 cameras. Systems of over three hundred cameras exist to try to reduce marker swap.

Vendors have constraint software to reduce problems from marker swapping since all markers appear identical. Unlike active marker systems and magnetic systems, passive systems do not require the user to wear wires or electronic equipment rather hundreds of rubber balls with reflective tape, which needs to be replaced periodically. The markers are usually attached directly to the skin, or they are velcroed to a performer wearing a full body spandex/lycra suit designed specifically for motion capture. This type of system can capture large numbers of markers at frame rates as high as 2000fps.

9. General discussion

Measurement of foot pressure distribution (FPD) is clinically useful because it can identify anatomical foot deformities , guide the diagnosis and treatment of gait disorders and falls, as well lead to strategies for preventing pressure ulcers in diabetes. Age-related anatomical and physiological changes in foot bone and ligament structure affect FPD during gait . Gait analysis of healthy elderly people has revealed decreased stride length, reduced step force and increased variability in gait parameters. These findings indicated that unsteadiness during walking is increased in the community-dwelling elderly people, posing a risk for falls. Age is independently associated with lower pressure under the heel, midfoot, and hallux in the multivariate analysis. Foot pressure studies during walking have focused on specific pathology and deformity specific anatomical areas, exercise and younger subjects. Knowledge of the plantar FPD map during normal walking in healthy elderly people is lacking. It is not known if distribution of plantar pressure, force, and load across several anatomical regions of the foot during walking is different between young and old.

The human foot plays an important role in both load support and shock absorption during walking. Shoes and insoles have been designed to protect the foot and facilitate proper foot functions for daily activities. An important determinant for a functional and comfortable foot support is how well it fits with the plantar foot shape. The foot shapes corresponding to different weight-bearing conditions are believed to be unique and can provide a more comprehensive description of the foot-insole interaction. It is important to understand the foot shape and its change under weight bearing and to determine which foot shape would best be adopted as the deciding factor in designing the support shape.

Previous studies on the anthropometrics of foot shape used varied protocols and measurement devices. Most approaches directly measure the foot length, breadth, height, and girth dimensions using sliding caliper, cloth tape, flat ruler, etc. These measurements may vary because of inconsistencies in positioning and the orientation of scales.

Benninghoff (1949) stated that the navicular bone was depressed, on average, 6.5 mm when bearing weight; the foot arch prolonged up to 19 mm within the second ray and 8 mm within the fifth ray upon weight bearing.

Carlsöö and Wetzenstein(1968) mentioned a quite different finding: that weight bearing caused no significant change in foot length and foot height. The different results found by these researchers may be due to an inconsistency in measuring positions, so that the actual foot joint orientation and amount of load undertaken were different.

Kayano (1986) used a surface-mounted electronic arch gauge to monitor the medial arch of the foot during normal walking. It was found that the medial arch length changed at different phases of gait. The degree of change in the length of the arch ranged from 3.7 to 9.5 mm. A similar method was used by Umeki (1991), who investigated the factors that influenced the length of the medial arch of the foot in normal adults under various passive motions and loads on the foot. It was found that the medial arch was lengthened and the foot was abducted when a vertical load was added to it. Shortening was observed when the first metatarsophalangeal joint was manually dorsi-flexed. The results indicated that the medial arch length would change with weight bearing and foot positioning. The use of skin-mounted measurement techniques may limit the accuracy of measuring the kinematics estimates of motion. This kind of error becomes considerable, as the foot shape alteration is relatively small.

Borchers et al. (1995) used a commercial light-striping laser digitizer to scan a foot in a non-weight-bearing condition and a 95-percent body-weight-bearing condition.

This kind of foot digitizing method avoided the error caused by skin displacement and tissue distortion. These shape variations gave the researchers ideas about the shape difference between a non-weight-bearing foot and a weight-bearing foot. Quantitative analyses and descriptions of these alterations are still limited.

Firstly, in clinical evaluation, foot type classification methods are based principally on morphology. It has been assumed that a given structural foot type will display certain functional characteristics and these, in turn, will be related to pathomechanics of the foot and the lower extremity. This kind of model assumes that function and kinematics can be assigned to a foot mainly based on its morphology. This is a fundamental but questionable assumption. One other model takes account of normal joint alignment and some functional components of foot mechanics. This classification is based on quantification of the frontal plane components of pronation. The focus of this approach is on neutral and resting (static) calcaneal stance position, subtalar joint range of motion and subtalar joint neutral position. Common to both is an attempt to predict dynamic foot function by using static measurements. However, recent reports have seriously questioned the reliability of clinical measurement of the criteria for definition of a normal foot and the validity of static measurements to predict dynamic foot functional behaviour.

Investigators have tried to evaluate the effect of foot type on the occurrence of injury during sporting activities. Ilahi and Kohl (1998) reviewed the English language literature from 1966 to 1997 to explore the scientific rationale for the clinical assumption that lower limb malalignment is a contributing factor in lower limb overuse injuries. They concluded that the literature generally did not support the clinical belief that decreased longitudinal arch and/or varus tibiofemoral alignment has a detrimental effect on the occurrence of injury. Results were frequently conflicting with dissimilar methodology, including outcome measures being considered as the principal reason for the diverse findings.

Factors other than intrinsic biomechanical abnormalities may also have a major role in the aetiology of sport injuries. These include extrinsic factors such as improper training techniques and weekly mileage, poor equipment, inappropriate shoes, unsuitable terrain, and other intrinsic factors such as bone geometry, previous injuries and years of running experience. The role of these factors (potential confounders) is frequently not taken into account in studies that have attempted to address the relationship between foot type and the occurrence of injury. Indeed, the multifactorial nature of running injuries makes it difficult to draw clear and sound conclusions on the specific aetiological factors contributing to a particular injury. Many injuries will be self-limiting and need no specific treatment. However, orthotic intervention may be appropriate in those injuries resulting from identifiable abnormal biomechanics.

The term 'foot orthosis' covers a wide spectrum of externally applied devices, ranging from simple arch supports to custom-made dynamic ankle-foot drop foot splints. The goal of orthotic prescription is variable, depending on specific need. However, functional orthoses are usually prescribed in an attempt to alter foot function with the expectation that they will guide the foot through the weight bearing stance phase of gait to promote overall biomechanical efficiency. Their use is somewhat empirical and frequently based on assumptions and insufficient clinical assessment.

Gait analysis systems, including motion-capturing devices, force platforms and foot pressure measurement systems, are sometimes employed to investigate the effect of foot type and orthotic application on foot biomechanics. These complex systems frequently use different calibration methods, and data collection, analysis and reduction strategies, complicating across-study comparisons. In routine clinical practice it is often common to use skin markers on the body to represent different segments during movement analysis. However, the validity of demonstrating movement of any skeletal segment by marker placement on skin has been always a matter of concern among clinicians and researchers. So called 'skin movement artefacts' introduce errors as a result of the relative movement between skin and underlying bone. However, a comparison between rotation measured with skin-mounted and bone-inserted markers showed that tibiocalcaneal rotation was generally well reproduced with external markers. Because of the invasive nature of bone markers, the study was limited to a small group and subsequently the study is of limited generalisability. Other noninvasive methods need to be employed in a larger population to investigate the difference between skeletal and soft tissue body segments.

Using bone-inserted markers, Stacoff et al. (2000) investigated the movement pattern of calcaneus and tibia during the stance phase of running in volunteers wearing shoes with and without orthoses. The results of the study were reported to indicate that tibiocalcaneal movement patterns were not substantially altered by medially placed foot orthoses, either anteriorly beneath the medial arch or posteriorly under the calcaneus. Differences between the volunteers were found to be significantly larger than between the orthotic conditions. Although all volunteers used the same running shoe and orthoses, both bone and shoe movements were interestingly found to be typical for each participant, indicating a participant-specific and unsystematic effect of orthotic intervention. The comparison of eversion velocities measured with skeletal and shoe markers showed a significant difference, emphasising a relative movement between the shoe and the foot.

The effect of shoes in altering the pattern of foot and ankle movement, and the lower extremity as a whole, is usually underestimated. In attempting to attenuate shock during walking and running, shoes may promote excessive movement. Shoe medial and lateral counter instability resulting from fracture or breakdown at midsole-counter interface could also produce an excessive range of motion. The common experimental practice of putting surface markers on shoes to represent the foot in different gait analysis techniques may introduce a further source of error, potentially complicating the comparison of the results from different studies.

Orthotic intervention is believed to influence the pattern of lower extremity movement through a combination of mechanical control and biofeedback. It has been speculated that orthoses placed under the midfoot and forefoot may increase the afferent feedback from cutaneous receptors, which may lead to reduced eversion due to muscular contraction of inverting muscles. More recently, the new concept of 'minimising muscle activity' has been proposed to explain the effect of applying shoe inserts and orthoses in sporting activities.

10. Conclusion

The human foot is a highly complex structure, with 26 major bones and more than 30 synovial joints. The foot plays a very important role in both load support and shock absorption during walking.This complexity of the foot is also regognisable, by the fact that there are so many different kinds of approaches for the analyzing of the foot shape and pressure distribution. Some of these methods are so complicated, expensive and time consuming to work with that are used only for research purposes. Because of the complexity of the foot most of these methods are analyzing only some aspects and lacking some others, so the good understanding of the foot biomechanics and kinematics is limited up to a point.

Of course with the rappid progress of technology, new methods have been developed and scientist are able to have a more sferical view of the human foot and better understanding of it.

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