Effect of Foot Orthoses on GRF in Running Gait

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Abstract

For many years foot orthoses have been used to treat injuries of the feet, lower limb and back. Much of the evidence for their use has been anecdotal and measurement of kinematic or kinetic effects has been inconclusive. A single subject was selected for this case study to test the effect of orthoses on ground reaction forces during running.

The subject was a competitive multi-sports athlete, and a heel strike runner (characterized as a runner who's heel is the first part of the foot to contact the ground).

The experiment was conducted in a hall on a 40m curved running track with a force plate on one side. Timing lights were placed 5m from each end of the plate to measure speed and a video camera recorded the foot strike on the plate. The subject was asked to run at constant speed while wearing shoes and shoes with foot orthoses, at two self-selected speeds. Data from left and right foot was combined for analysis.

The results showed a significant decrease in the magnitude of the vertical impact peak and the maximum vertical peak while the time to vertical impact peak was increased when wearing foot orthoses. Significant reductions were also seen in the peak posterior shear with both the time to peak and magnitude of the peak being changed by wearing foot orthoses.

The mediolateral force was characterized by a medial impact followed by larger lateral impulse. It is the lateral force in the absorption phase of stance that is responsible for pronation, however no changes were seen in the mediolateral ground reaction force with the use of foot orthoses. This indicates that there is no acute effect in the shear forces that act at approximately right angles to the subtalar joint axis. If orthoses have an acute effect on the lower limb it is likely to be complex and highly patient specific.

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Acknowledgments

There are a couple of people I would like to acknowledge for their help and support.

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Glossary

Abduction – A frontal plane motion where the segment rotates away from the mid line.

Adduction – A frontal plane notion where the segment rotates towards the mid line.

Ankle - The articulation of the tibia and talus (Talocrural joint).

Dorsiflexion – Flexion of the talocrural joint.

Eversion – Motion occurring in the frontal plane where the plantar aspect of the foot is tilted away from the mid line of the body, about axes in the sagittal and transverse planes.

Inversion – Motion occurring in the frontal plane where the plantar aspect of the foot is tilted towards the mid line of the body, about axes in the sagittal and transverse planes.

Foot orthoses – Orthopedic appliances used to correct deformity or inadequacy of the foot and lower limb. Also referred to as Orthotics.

Plantarflexion – Extension of the talocrural joint.

Pronation – A complex motion of the rear foot that requires movement in all three anatomical planes.

Midtarsal joint – Articulation between the calcaneus and cuboid and the talus and navicular.

Rear foot valgus – An everted structural position of the rear foot.

Rear foot varus – An inverted structural position of the rear foot.

Subtalar joint – Articulation between the talus and the calcaneus.

Valgus – The distal segment is angled away from the mid line of the body.

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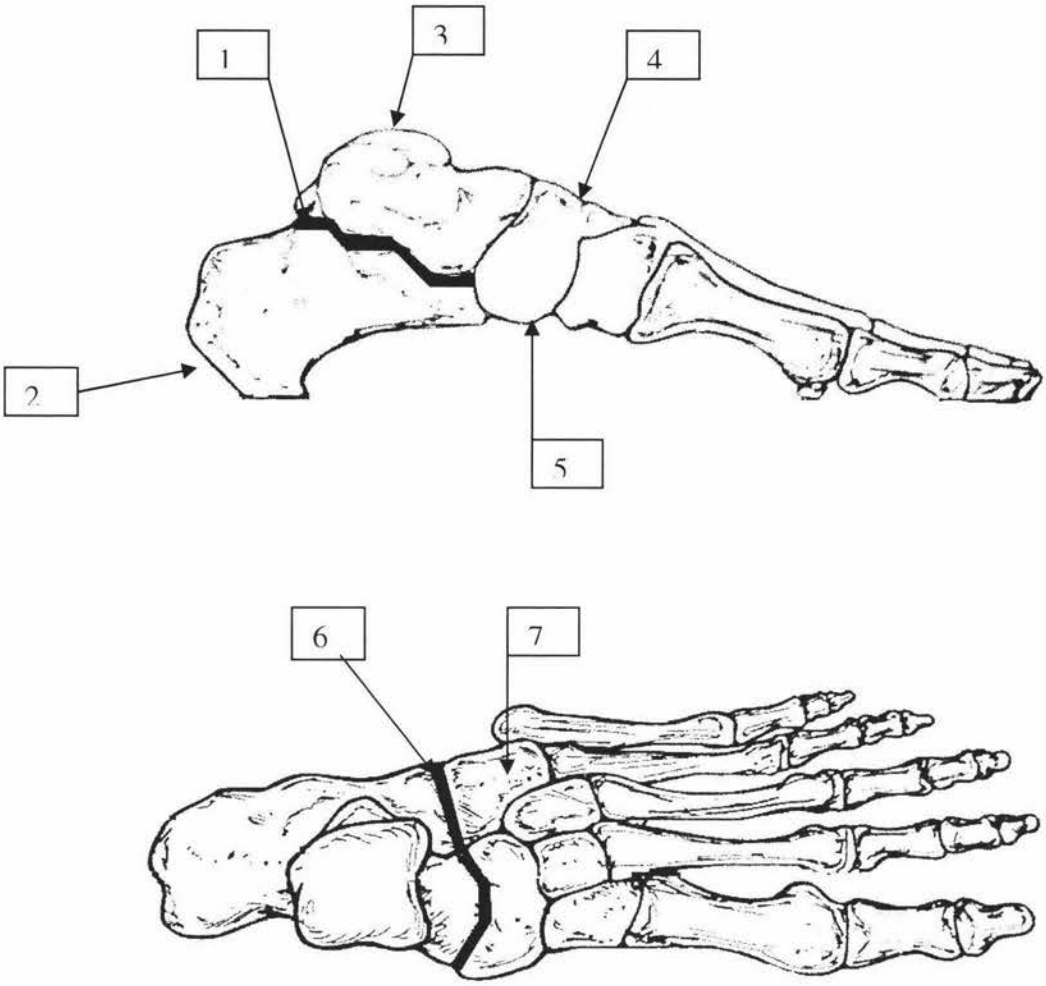

Figure 1 Selected Joints and Bones of the Foot

Key

1. Subtalar Joint
2. Calcaneus
3. Talus
4. Navicular
5. Navicular tuberosity
6. Midtarsal Joint
7. **Cuboid**

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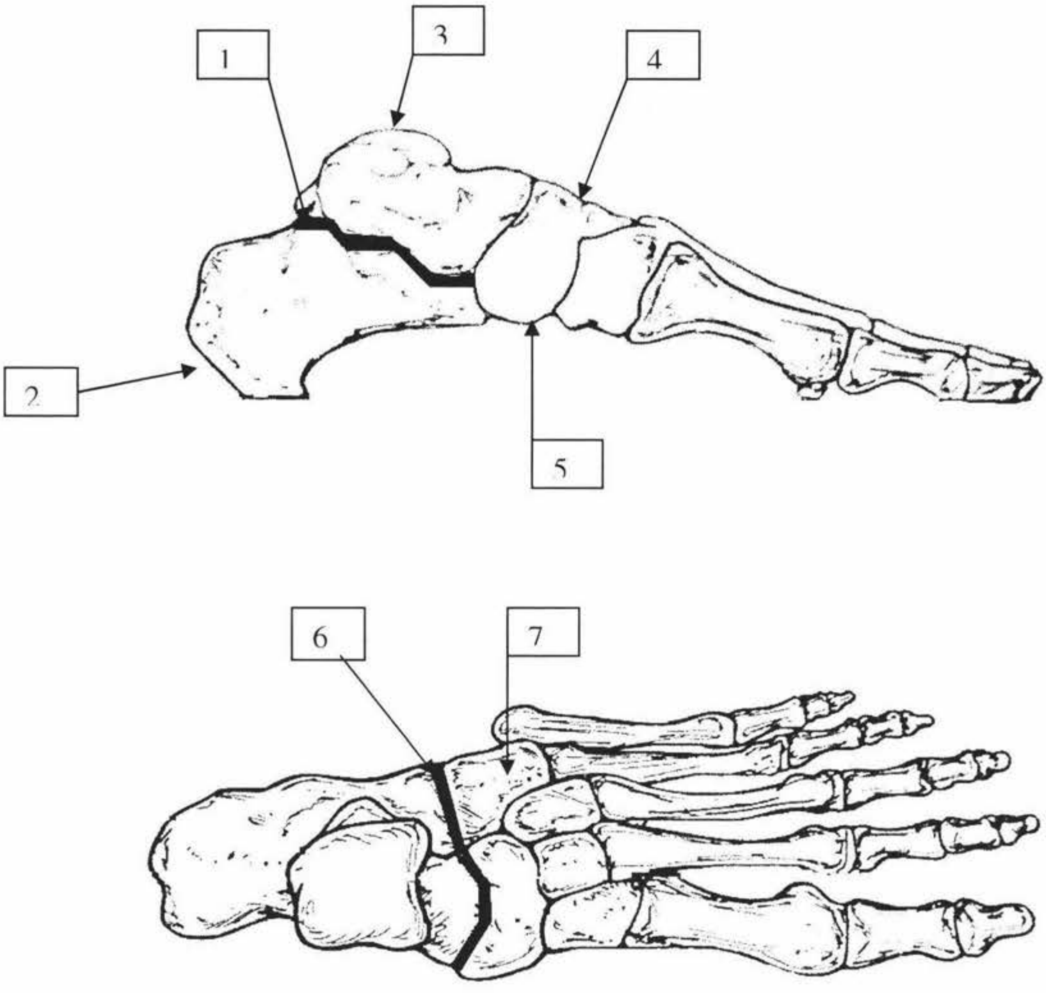

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Key

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2. Calcaneus

3. Talus

4. Navicular

5. Navicular tuberosity

6. Midtarsal Joint

7. Cuboid

Introduction

For many years foot orthoses of various types have been given to athletes as a remedy for a wide range of injuries. Indeed, not only injuries of the leg and foot have been treated with foot orthoses but also injuries as remote from the foot as the lower back. Clinical texts have advocated the use of orthoses for specific injuries in order to control the abnormal mechanics that have been speculated to cause tissue damage at the site of the injury, (Valmassy, 1996; Brukner & Khan, 1998; Subotnick, 2001).

Foot orthoses may be able to help resolve a number of injuries, such as overuse injuries, in which abnormal biomechanical function may be a significant etiological factor. Research has shown an interaction between the foot, the lower limb, and the pelvis. This coupled system will affect bone, articular cartilage, ligaments, and muscles that are inside this chain. Consequently, direct control of the foot by an orthotic might be expected to affect the more proximal links in the chain.

Clinical purpose of foot orthoses

Philps (1990) has suggested that the function of a foot orthotic when placed under the foot is to synchronize the mechanics of the lower limb. This is done by holding the foot in its optimal functional position, with the subtalar joint in a neutral position in the mid-stance phase of gait. This position is currently considered necessary for normal function by many clinicians.

Valmassy (1996) commented that orthoses are prescriptive medical devices, which are used to lend assistance to the realignment of the lower limb joints. Clinically podiatrists assume that orthoses decrease the amount of abnormal stress in the lower limb, which may be caused by poor joint alignment and muscle function.

Valmassy & Subotnick (1999) commented that functional foot orthoses are prescribed to guide the foot through the stance phase of gait, which will in turn promote mechanical efficiency.

Valmassy (1996) lists the following specific kinematic changes that could be expected with the introduction of functional foot orthoses.

Normalise ankle dorsiflexion and plantar flexion.

Produce normal knee flexion at heel contact to improve shock absorption.

Give proper hip flexion/extension

Create efficient internal/external lower extremity motion.

Produce normal subtalar joint/midtarsal joint pronation/supination.

This list shows that the author has a high expectation that significant effects will be seen throughout the kinematic chain from the distal to the proximal segment.
In contrast Brukner & Khan (1998), suggested that orthoses act to control excessive subtalar joint and mid-tarsal joint movements that may occur to compensate for structural abnormalities. These comments lead to two very important questions that are not easily answered, but will provide some insight in to the use of orthoses. Firstly, despite the constant references in clinical texts to normal, proper, and abnormal function, it has been difficult to conclude exactly what is normal and proper when it comes to foot function. Pronation is a complex multi plane motion that involves the subtalar joint, mid-tarsal joint, and ankle. The motion at each of these joints is dictated

by its specific axis of motion. Anatomical studies are beginning to show that the axes of the subtalar and mid-tarsal joints are not identical in all feet. Valmassy (1996) suggested that a high axis of subtalar joint motion will result in more tibial rotation while a low axis will result in a reduced range of motion in the tibia. Therefore, because the link between the foot and lower limb is subject-specific it is hard to predict joint motion between individuals, (See Figure 2).

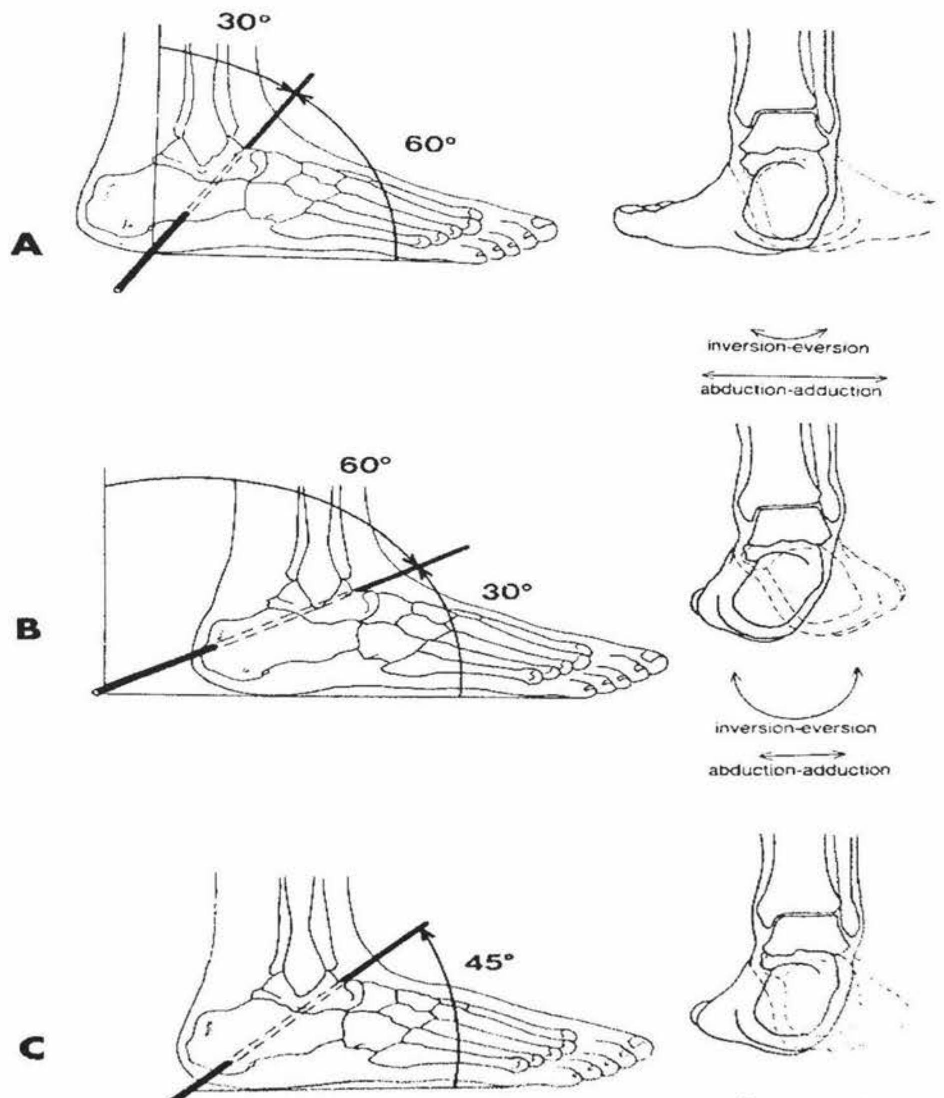


Figure 2 Variation in the Subtalar Joint Axis of Motion. Reproduced from, Valmassy (1996); Clinical biomechanics of the lower extremities, Mosby.

The second question that could be raised in relation to orthotic control of the lower limb is what segment drives the rotation? Belchamber & van den Bogert (2000), while investigating power flow in running found that, in general, internal rotation of the tibia produced pronation in the foot but, in some subjects for a brief period (between 40 & 60% of stance), the foot was producing rotation of the tibia. This indicates that there maybe a point in the gait cycle at which an orthosis could influence the leg and thigh. The study also showed that subject specific function was a strong and significant influence in the study.

Anatomy and motion of the ankle joint

When foot orthoses are used to control either the foot or leg, the foot is being used to create altered function. Foot orthoses are believed to control primarily the function of the subtalar joint, which might alter function in segments that are connected to its motion. The subtalar joint is an articulation of the talus and calcaneus with its axis of motion orientated approximately 42° from a transverse plane and approximately 16° from a sagittal plane, (See Figure 3). Because of this oblique axis, pronation produces motion at this joint that is multi-plane and, in a closed kinetic chain this will influence the motion of the more distal mid-tarsal joint and the more proximal ankle, knee and hip joints.

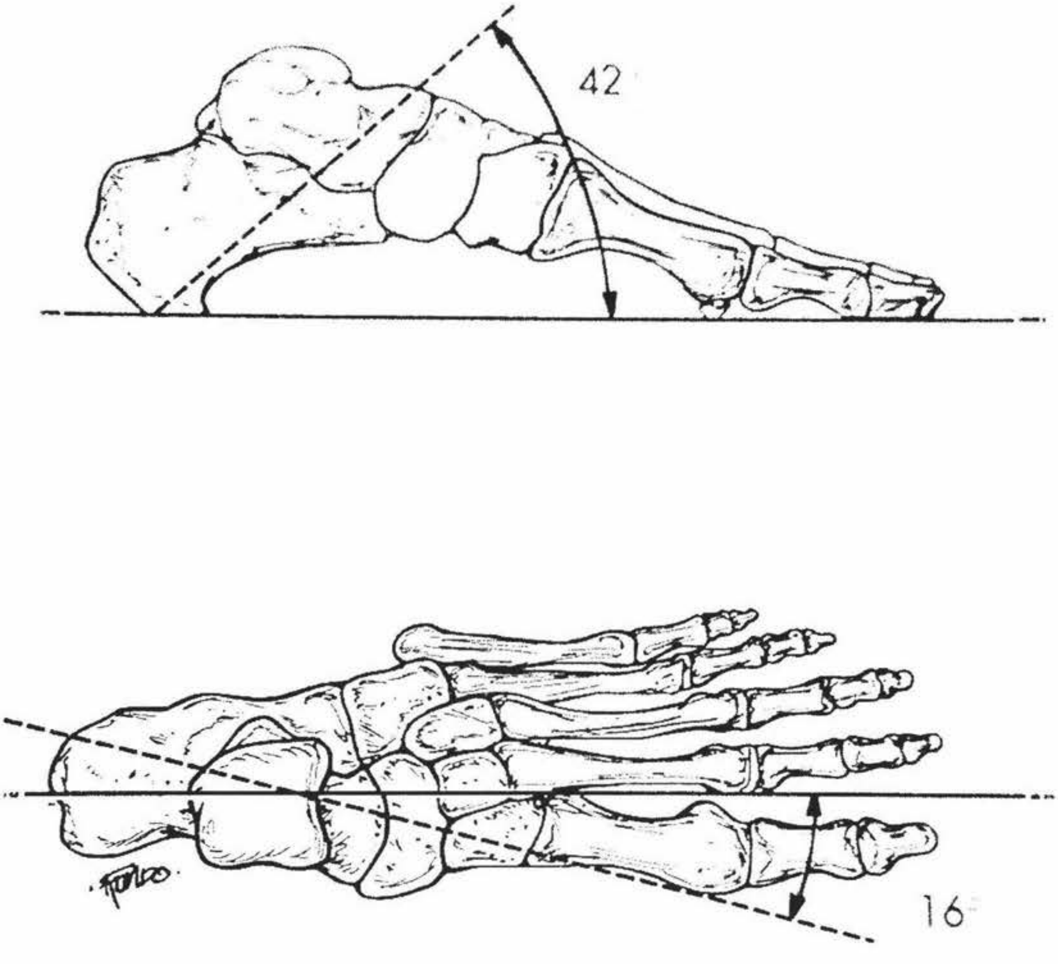


Figure 3 Subtalar Joint Axes of Motion. Reproduced from, Valmassy (1996); Clinical biomechanics of the lower extremities, Mosby.

In a closed kinetic chain, clinically podiatrists expect that rotation about the subtalar joint axis is associated with the following motion in the foot.

Pronation	Calcaneal eversion
	Talar adduction
	Talar plantarflexion
Supination	Calcaneal inversion
	Talar abduction
	Talar dorsiflexion

The subtalar joint allows motion in the rear foot but it requires the more distal midtarsal joint to compensate for its rotation in the frontal plane to maintain ground contact of the forefoot.

The midtarsal joint is made up of the combined articulations of the calcaneocuboid and talonavicular joints. The midtarsal joint has an oblique and longitudinal axis of motion, (See Figure 4).

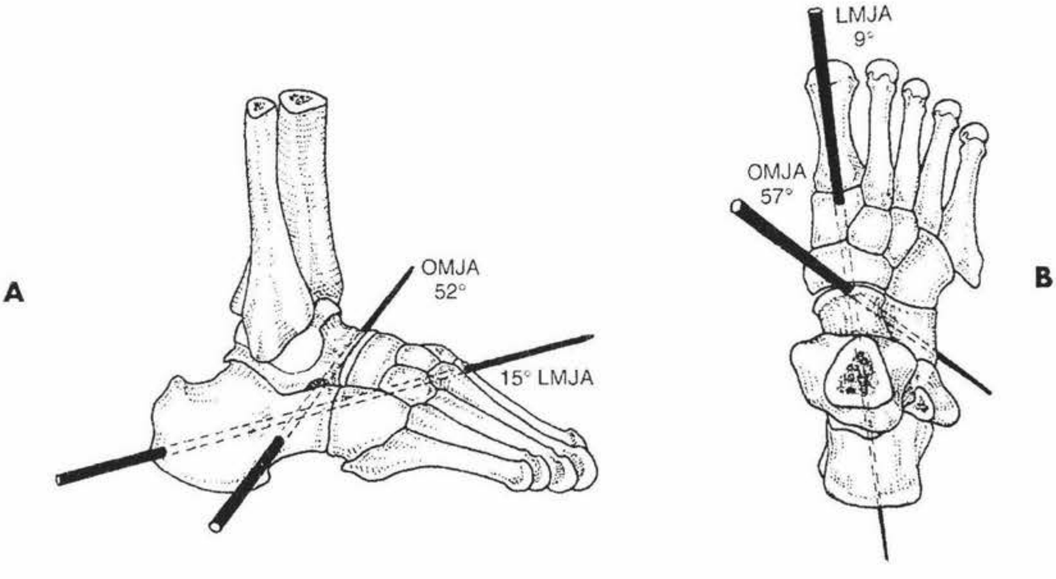


Figure 4 Midtarsal Joint Axes of Motion. Reproduced from, Valmassy (1996); Clinical biomechanics of the lower extremities, Mosby.

The primary motion in the longitudinal axis is inversion and eversion while the oblique axis allows dorsiflexion coupled with abduction and plantarflexion coupled with adduction, (See Figure 4).

The motion of the subtalar and midtarsal joints is linked to allow the foot to maintain total plantar contact during gait. As the rear foot is everting the forefoot is inverting and vice-versa, and it is the combination of these movements that foot orthoses hope to control.

A medial wedged foot orthotic is believed to increases pressure under the medial border of the calcaneus. The medial wedge, which extends from the back of the heel to a point under the sustentaculum tali is used to control rear foot function. Clinical prescribers of foot orthoses propose that this wedge will either decrease the magnitude of rear foot eversion or slow the velocity of rear foot eversion. In addition to the rear foot wedge the solid shell may act as a brace to stabilize the mid foot, (See Figure 5). This control at the subtalar and midtarsal joints is speculated to have an effect on tibial rotation and rotation of the rest of the lower limb.

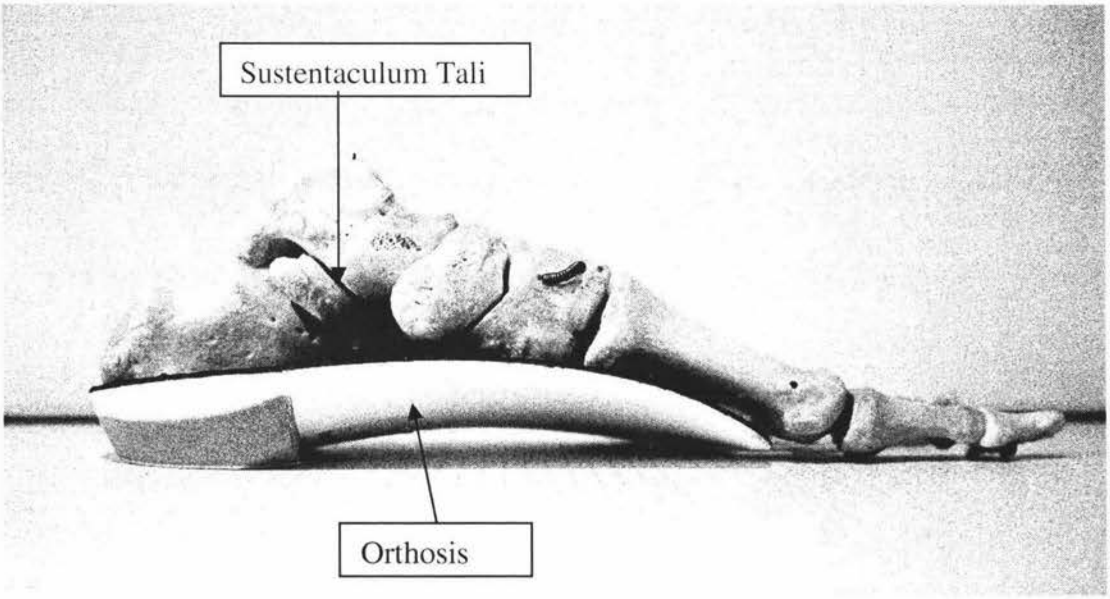


Figure 5 Medial view of foot positioned on orthosis

Lower Limb motion during running

During heel-toe running the heel contacts the ground on the lateral boarder with the subtalar joint in mild varus. The foot rotates to the support surface by plantar flexion at the ankle. While the ankle is plantarflexing, the subtalar joint is pronating, which will allow the foot to become a loose adaptor for weight acceptance. The foot can become a loose adaptor because subtalar joint pronation unlocks the midtarsal joint, which allows the foot to conform to the ground. This unlocking also produces a compliant arch, which as it increases radius of curvature absorbs shock by storing energy in the ligaments of the mid foot. Intrinsic and extrinsic muscles associated with foot function will dissipate energy at this time. The sub-talar joint continues to pronate and reaches its peak pronation between 35% and 45% of the stance phase of gait in the normal runner. During this time, the tibia and femur are rotating internally and the knee is flexing, (See Figure 6). This part of running gait is called the absorption phase because this is the period of stance where the energy of impact is absorbed. During this period of the stance phase of gait pronation of the subtalar joint may be altered. The gait cycle now becomes propulsive and the foot begins to supinate reaching a neutral position at about 70% of the stance phase. This supination locks the midtarsal joint by rotating the calcaneocuboid joint into a close packed position, which produces a rigid foot for effective propulsive. Through the propulsion phase the ankle plantarflexes and the tibia and femur are rotating externally, (See Figure 6). Excessive pronation that continues during the propulsive phase of gait has been speculated to interfere with rotation of the tibia and femur (Hamill & Knutzen,1995; Lafortune et al, 2000).

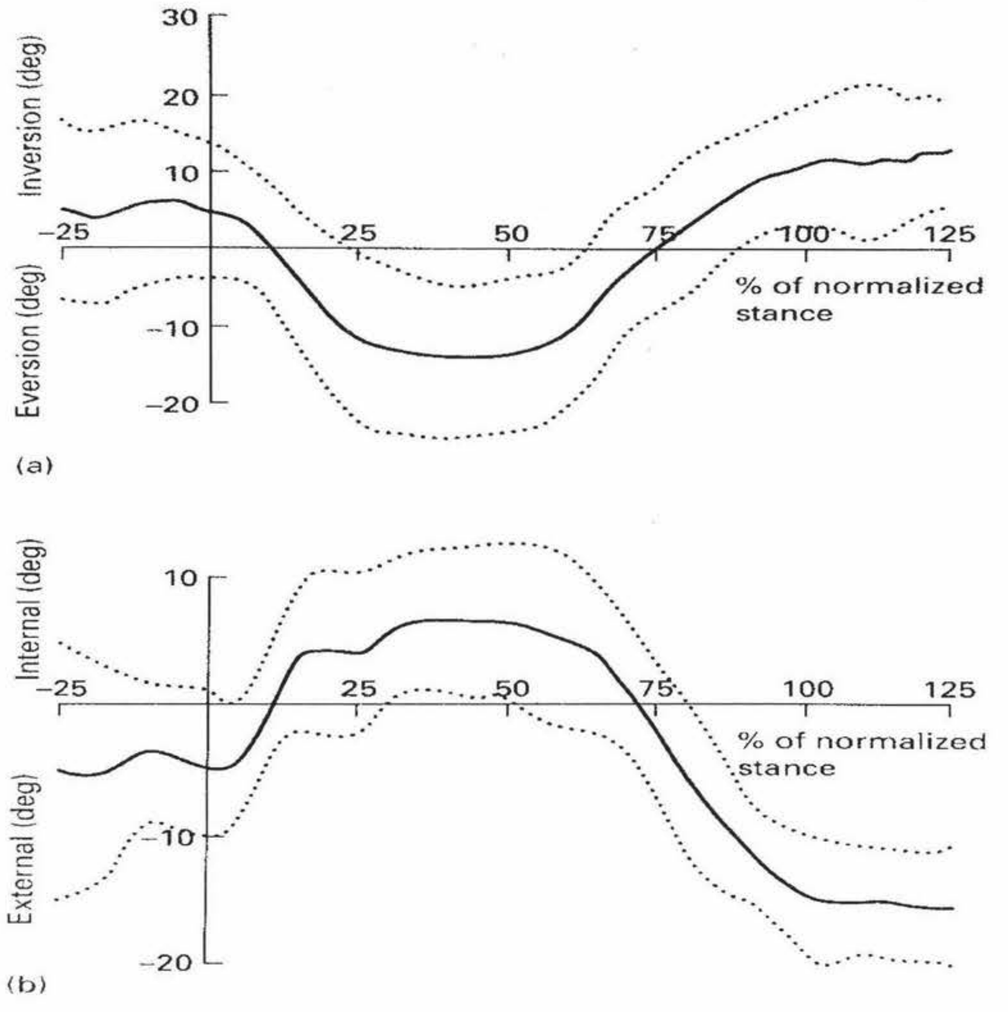


Figure 6 Rotation of sub-talar joint (a) and tibia (b) during the stance phase of running. Reproduced from, Lafortune, Valiant, Mc Lean, (2000); Running, Blackwell Science.

During the stance phase of running the foot is exposed to ground reaction forces in the vertical, mediolateral and anteroposterior directions. These forces will have a marked effect on the motion of the foot and lower limb. At heel strike, a vertical impact peak occupies approximately the first 20 – 30ms of stance. This phase of absorption is referred to as passive because it is not controlled by muscle activity and it is thought that passive structures are used to absorb energy. In the normal heel-toe runner this vertical impact force is applied lateral to the subtalar joint axis and posterior to the ankle joint axis, and results in a pronatory moment at the subtalar joint. During the initial contact phase of gait the ground reaction force has medial and posterior components as the shoe is stopped by the ground.

After the passive phase active absorption starts as muscles become the primary controllers of ground reaction forces. During the foot flat phase of gait a substantial peak in vertical ground reaction force is seen and the mediolateral force is in a lateral direction. At this phase of stance the centre of pressure is lateral to the subtalar joint which when combined with the lateral shear force produces a pronatory moment at the subtalar joint. When the anteroposterior force reverses and becomes propulsive, the vertical force reduces and the mediolateral force also reverses to a medial direction. The center of pressure now moves to the medial side of the subtalar joint axis, which produces a supinatory moment at the subtalar joint (Hamill & Knutzen, 1995).

Previous Clinical Research

Many studies have been conducted in an attempt to shed light on the anecdotal reports of clinical benefits of foot orthoses that pervade the clinical literature. No general consensus has been reached at this stage, possibly because of limitations in study design, diagnosis and, most importantly, the uncontrolled effect of individual variation.

Williams, McClay & Hamill (2001), studied 40 male and female runners, with an average age of 27years in a non-randomized 2 group injury survey to look at injury in high and low arched feet. They found that subjects with low arches tended to suffer from medial injuries of a soft tissue type, whereas subjects with high arched feet tended toward lateral and bony types of injury. This leads to the conclusion that, in lower arched feet, over work of muscles and ligaments is likely to cause more soft tissue injury, while in high arched feet, impact loads are higher and lead to shock loading types of injury in bone.

Bennett et al (2001) studied a group of 15 year old high school track runners of whom 15 had medial tibial stress syndrome and 21 were injury free. They saw an increased rate of medial tibial stress syndrome in subjects with higher navicular drop measurements. The navicular drop test is performed by measuring the amount that the navicular drops in relation to the ground when going from standing on both feet to one foot. The injured group had a navicular drop of 6.8 mm while the control group had a navicular drop of 3.6 mm. It was suggested that a pronated foot type is related to medial tibial stress syndrome.

London (2002), also used navicular drop when studying a group women, aged between the ages of 25 and 65 who had enrolled in a 10 week exercise training program. It was noted that in those subjects exhibiting a range of injuries such as patellofemoral dysfunction, plantar fasciitis, Achilles tendonitis and hamstring strains the mean navicular drop measurement was 8.3mm compared with a mean of 6.2mm in the uninjured participants. The clinical significance of this is uncertain because of the small difference navicular drop (2.1mm).

Razeghi & Batt (2000), in a review article reported that Messier & Pithala (1998) used rear foot motion in a study of etiologic factors in running injuries. They suggested that a higher range of heel motion was seen in people with shin pain, planter fasciitis and iliotibial band syndrome, whereas people who were uninjured were claimed to be in a normal range.

In contrast Nawoczenski et al (1998), studied 20 male and female recreational runners who were radiographically placed into a high or low arched group. The low arched group had a mean age of 28 years while the high arched group had a mean age of 31 years. They showed that in the period from heel strike to maximum pronation, similar eversion maxima occurred in both high and low arches, suggesting that the pronation that orthoses are designed to control may be a poor indicator of foot function.

Morag & Cavenagh (1999) studied 55 men and women between the ages of 20 and 70 years in an attempt to show variations in kinematic and kinetic variables based on anthropometric measurements. The authors concluded that high plantar pressures would be expected in the mid foot of flat-footed subjects, whereas higher plantar pressures would be expected in the heels and first metatarsal phalangeal joint in higher arched subjects. In spite of the lack of direct evidence to prove a link between foot function and injury, many clinical texts claim a clinical association between overuse injury and excessive pronation, (Brukner & Khan,1998; Subotnick, 2001; Valmassy, 1996).

Clinical studies into the effectiveness of foot orthoses as a treatment for injury of the lower limb have also shown highly variable rates of success.
Razeghi & Batt (2000) reported that Gross, Davlin & Evanski (1991), conducted a study into the effect of orthoses on injuries in a group of middle distance runners. They saw an improvement in symptoms in some subjects but 24% showed no benefit and 13% showed an increase in symptoms. When an increase in symptom was recorded patients in the study suggested that the insoles were poorly fitted or diagnosis was incorrect. However, this may not be the case, and may indicate that foot orthoses may exacerbate problems in some people.

Tis et al (2000), used a survey style study to follow the progress of 15 middle distance runners, each of whom wore foot orthoses which where fitted for a range of running related injuries. The subjects were competitive trained athletes with an average age of 20.8 years. The authors stated that 7 of the 15 had a complete recovery after 13 months of using foot orthoses. This relatively low recovery rate (below 50%), reported after 13 months is by most clinical criteria, an inappropriate time frame in which to control symptoms of injury. These people may have just got better with time.

Gross et al (2002) conducted a pain questionaire study to assess the effect of semi-rigid orthoses on pain and disability of 15 male and female walking patients suffering from plantar fasciitis. The mean age of the subjects was 44.7 years and they had suffered the symptoms of plantar fasciitis for an average 21 months. The subjects were assessed by a pre and post orthotic foot function index and a pain rating. It was concluded that, on average, foot orthoses did reduce the pain of plantar fasciitis during walking. This study did not indicate a permanent cure for this condition, simply a reduced level of pain. This study did not have a control group and some subjects were asked to improve their shoes, which may have confounded the results.

Previous Experimental Research

Clinical benefits from the use of orthoses can be seen when reviewing case studies and clinical research. What is still unclear is how foot orthoses reduce symptoms related to musculo-skeletal injury. Foot orthoses are speculated to reduce the range of motion in the subtalar joint and midtarsal joint during walking and running. In a review paper Razeghi & Batt (2000), identified a number of kinematic variables that might be significantly altered by foot orthoses. This author feels that these variables should be those used to indicate a treatment effect. The factors that have been studied relate to the rear foot and its function, such as maximum pronation, pronation velocity, and time to total pronation.

Brown et al (1995), in a study of 24 male and female over pronated walking subjects measured pronation while wearing shoes, shoes with soft insoles, and shoes with biomechanical foot orthoses. Their results showed that no change could be seen in maximum pronation, calcaneal eversion and total pronation in any condition. However, they did find that time to total pronation was greater with the biomechanical orthoses. This shows a reduced velocity of pronation but the amount of pronation stays the same.

Nowoczenski, Cook & Saltzman (1995), used a more sophisticated 3D kinematic study to observe kinematic changes in the motion of leg and rear foot during running. The study included 20 male and female recreational runners with a mean age of 28 years for the low arched group and 31 years for the high arched group. The authors concluded that orthoses had no effect on tibial adduction or frontal plane subtalar joint motion. However, a significant reduction was seen in maximum internal tibial rotation following heel strike. The authors suggested that this showed that the effect of foot orthoses occurs in the first 50% of stance. This would seem reasonable because after mid stance, the heel lifts and the orthosis has no further contact with the support surface.

In contrast, Mundermann et al (2003), studied the effect of rear foot wedges, molding, and wedged molded insoles on running gait of 21 male and female recreational runners. The average age of the subject group was 25.4 years and all were classified as over pronators who had never worn foot orthoses. Results showed that simple wedges designed to place the calcaneus in an inverted position reduced maximum foot eversion and ankle inversion moments. When molding was used, no reduction was seen in rear foot function in the first 50% of stance. However, she noted an increase in maximum foot inversion and maximum external rotation moments at the knee with molded and wedged insoles. This occurred in the last 50% of stance as the subtalar joint is naturally reducing pronation in preparation for push off. However in Mundermanns's study the medial wedges extended to the forefoot, which produce a re-supinatory effect not possible with standard foot orthoses because they only have a rear foot wedge.

Stacoff et al (2000) examined the effect of medial wedges on foot and lower limb function while running. The study used a 3D motion analysis system to track the motion of bone pins surgically placed into the feet and legs of 5 subjects. The subjects used in this study were not considered over pronators and had an average age of 28.6 years This study produced highly accurate results, as all skin movement artifact was removed, but found highly variable results from subject to subject. They concluded that a non-significant 2° change in total rear foot eversion was seen while running with medial wedges. They also reported that total eversion velocity was altered by 1.6 – 10°/s, which was not significant. It was concluded that the orthotic effect on rear foot eversion and tibial rotation was small and unsystematic over all the subjects.

Researchers have also looked at kinetic variable changes with foot orthotic intervention. The most common kinetic variables studied are ground reaction forces. Changes in ground reaction forces would be expected to significantly change the function of the lower limb because kinematics are driven by forces.

Morarty & Agosta (1998) studied 10 athletes while running with and without foot orthoses. They saw statistically significant changes in only 2 of the eight ground reaction force variables that they analyzed. The initial impact peak was reduced and the time to peak vertical force was significantly increased. Morarty & Agosta's(1998) study showed changes to impact loading, which may indicate that in the first 20 – 35ms of ground contact shock absorption is being reduced. This maybe a result of a reduced rate of pronation, which maybe necessary to absorb shock. However Hennig & Milani (1995) reported from a study by Edington, Fredrick & Cavanagh (1990), that during the initial impact peak over the first 35ms a substantial amount of rear foot varus still existed indicating that little subtalar joint pronation had occurred to alleviate this shock.

At present, it appears that the heel fat pad is the primary impact attenuating structure. Winter et al (1995) commented that, while walking, a large amount of energy was absorbed mainly in the heel pad and much less was absorbed by active eccentric contraction in the inverter and dorsiflexors muscles. This is further confirmed by a study by Christina et al (2001), which measured changes in vertical ground reaction forces while running before and after fatigue of the dorsiflexors and invertors of the foot. The study included 11male and female recreational runners with an average age of 24.3 years. The authors found no change in loading rates and maximum impact force with inverter fatigue, (Christina, Scott & Gilchrist, 2001).

When Mundermann et al (2003), applied medial rear foot wedges to the shoes of runners she saw a significant decrease in rear foot eversion and an increase in loading rates and maximum load. However, when molded wedged insoles were used in the same study loading rates where reduced and pronation increased. The decrease in impact loading rate and maximum load may have been a consequence of increased pronation, or possibly a result of the cupping around the heel preventing the fat pad under the heel from spreading laterally and, hence, increasing the vertical stiffness of the fat pad. This could be of benefit as the fat pad comes close to its mechanical limits during running.

Purpose of Study

The current body of knowledge has examined the effect of a wide range of foot control devices including simple arch supports, forefoot and rear foot wedges, soft and hard non-custom insoles, and customized functional foot orthoses. This Thesis will examine the effect of foot orthoses with a 4 degree rear foot wedge and a forefoot that is angled 2 degrees from the support surface, on ground reaction force during the absorption phase of running gait. The type of foot orthoses used in this study is the style currently used in podiatry practices in New Zealand. By studying the effect of this type of device some insight maybe obtained into the clinical practice of orthotic prescription.

In past research, attention has focused on vertical ground reaction force and to a lesser extent the anteroposterior and mediolateral ground reaction force when looking for orthotic effects. This study examines the acute effects of foot orthoses on ground reaction forces during the absorption phase of gait while running 2 different speeds. This research will focus on the mediolateral shear in an attempt to measure the forces that act at almost right angles to the axis of motion of the subtalar joint in addition to the vertical and anteroposterior ground reaction forces. Because foot orthoses are used clinically to reduce maximum pronation and the rate of pronation, changes should be seen in the mediolateral shear with their use.

Two running speeds will be used for this study because as running speed increases, so does the magnitude of the vertical loading rate. This indicates that mediolateral loading rates may also increase which may help in detection of small changes in this force.

Methodology

A case study design was chosen for this research project; one subject was selected and he acted as his own control during the study. This design was chosen because:

Substantial inter subject variability of kinematic parameters has been observed in orthotic intervention studies;

Ground reaction forces exhibit significant variability during running with the mediolateral force being the most variable both within and between subjects;

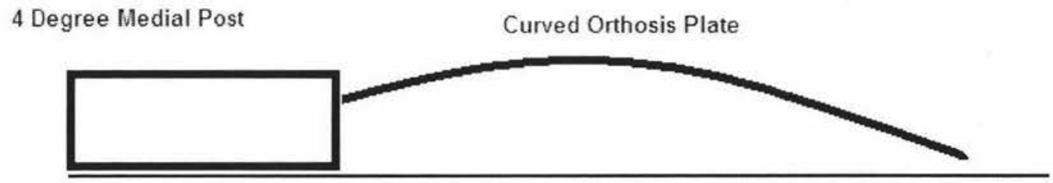
The between subject variability would tend to obscure any effect present in an individual.

Ethical approval for this research was granted by Massey universities ethics committee.

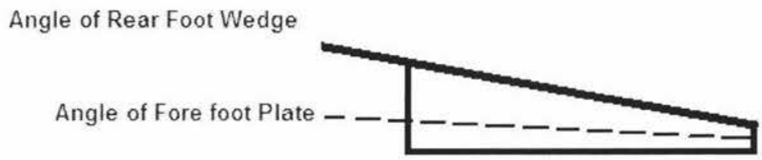
Orthoses

The orthoses used in this study were foot orthoses, commonly used by podiatrists in New Zealand to control pronation of the foot. This type of orthosis is made from a 3mm polypropylene shell that has been molded from a modified cast of the foot. The shell is proportioned to end behind the ball of the foot and caps up around the heel.

Each orthosis was fitted with a 4° rear foot medial wedge, which was constructed from EVA. The fore foot section of the shell was not posted, (medially wedged), and was parallel with the support surface minus 2°, (see Figure 7). This posting style will position the rear foot in 4° of varus while not controlling the forefoot. The shell was covered with non-shock absorbing vinyl, which allowed for ease of cleaning if necessary. The finished foot orthoses are shown in Figure 8.



Medial View of Orthosis



Frontal View of Orthosis

Figure 7 Schematic diagram of rearfoot and forefoot posting.

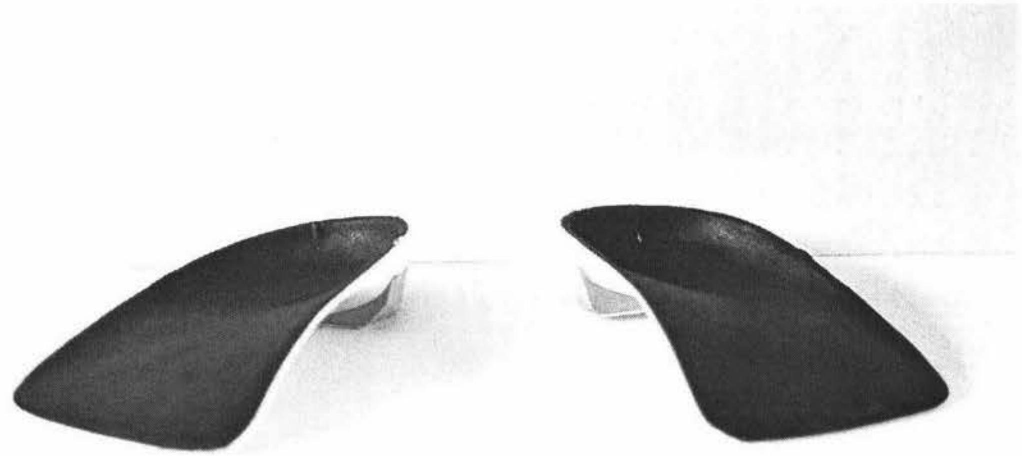


Figure 8 Foot orthoses used for study

The foot orthoses were designed to control the rear foot with the main loading area being the medial plantar heel and the area around the sustentaculum tali. The rear foot cup would help stop the foot slipping laterally off the shell of the foot orthoses.

The orthoses were checked for fit on the foot and in the shoe. It was important that the arch was not uncomfortably high, the shell was not past the break line of the foot and the heel cup was not too tight, (see Figure 9). When in the shoe the orthoses sat level on the shoe inner liner and were not tilted or forward of the inside of the heel counter of the shoe. Correct fitting of the orthoses in the shoe was important to ensure the position of the foot in the shoe was not changed by the foot orthoses and to maximize any change in foot function.

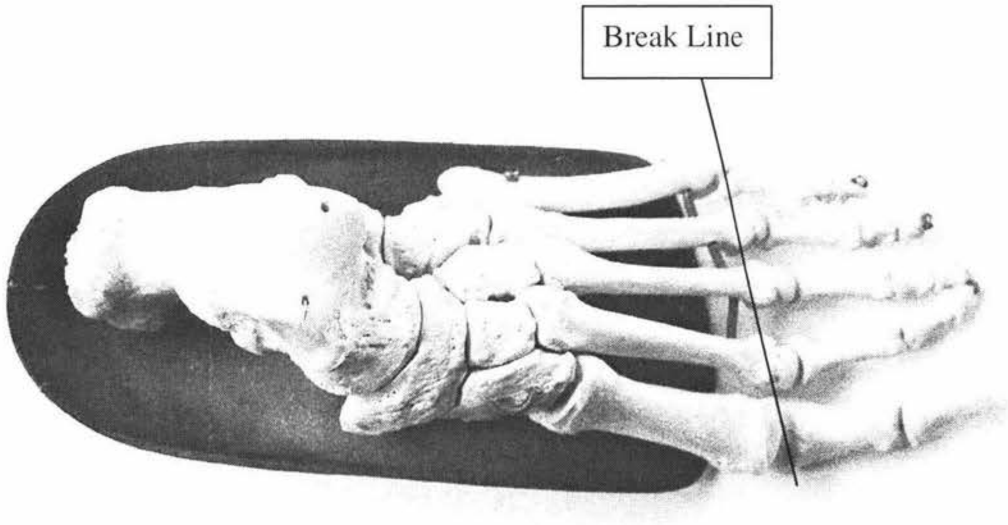


Figure 9 Foot positioned on orthoses

Subject

The subject selected for this case study was a well-trained competitive multi-sport athlete. The subject was 26 years old, measured 1.80m tall and weighed 725 N (74 kg). The proposed research was explained to the subject and his signed consent was given.

The subject was accustomed to running, which was considered important because it was expected that a trained runner would be able to maintain a consistent running speed and style through successive trials.

The subject had no injuries at the time of this study and was currently training and competing regularly.

For this study we required a runner that had a heel toe running style and did not wear foot control insoles of any type. The subject had a range of motion of more than 5° ankle dorsiflexion at the ankle and less than 0.5cm leg length discrepancy. For this study we did not want an over or under pronated foot, and so a subject was selected to have a relaxed calcaneal position less than 4° valgus.

The subject was required to have shoes that were not older than 8 months. Shoes older than this could be excessively worn and distorted which could alter foot function significantly. The shoes used during data collection were Asics 2090's.

Set Up

An indoor recreation hall was used for this study to allow for control of possible confounding factors during data collection. Bad weather, such as rain and wind, could alter normal running style substantially, and could prevent the study from being conducted at all. The portable force plate (AMTI Accugait) was positioned on a level surface to prevent rocking during foot strikes. Solid ramps were used either side of the force plate because the removable force plate was 6cm high and could not be recessed into the running area. The ramp was level with the force plate surface and extended each side for a distance of 5m. This gave a ramp slope of 1° each side, which was comfortable and appeared to create no disturbance to running style while in the data collection area. The ramp was covered with a non-cushioned carpet to help visually blend the ramp and force plate together to prevent direct targeting. The ramp and carpet were firmly attached to the floor with double sided tape to stop any slipping of either the carpet on the ramp and the ramp on the floor. The AMTI force plate was portable, which allowed it to be easily located and set up in the hall. The plate measured 50cm by 50cm and was 6cm high. The sampling rate was 200hz and a rising edge trigger was set at 20N on the vertical force channel. The force plate recorded vertical (Fz), anteroposterior (Fy), mediolateral (Fx) ground reaction forces and these data were collected on a laptop computer, (See Figure 10-12).

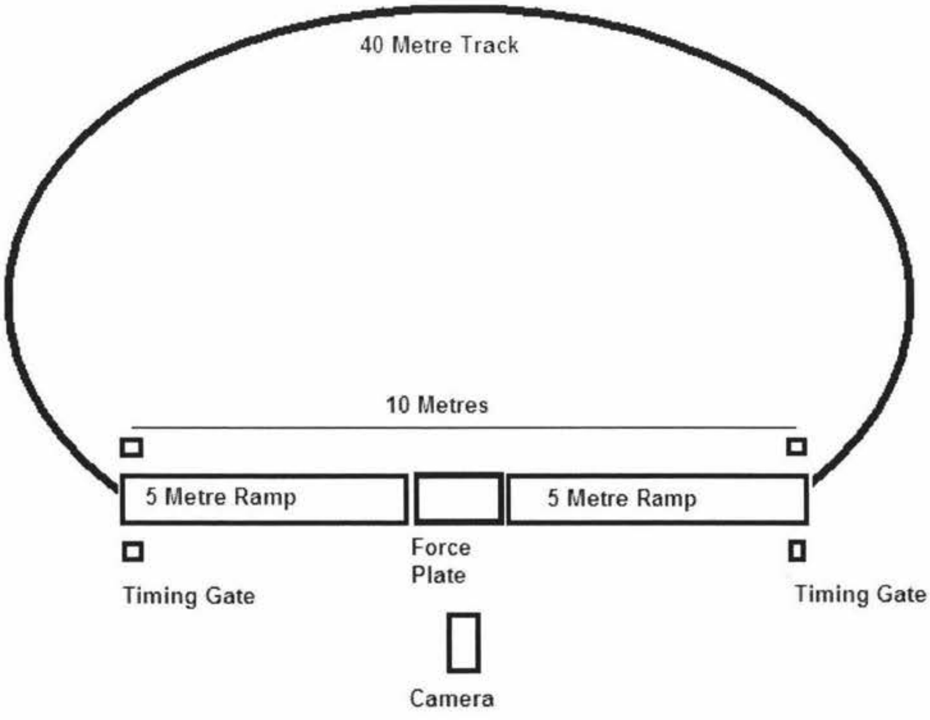


Figure 10 Schematic diagram of experiment set up.

Figure 11 Data Collection Set Up showing 40m circuit

Figure 12 Data Collection Set Up showing the ramp and force plate.

A 40m running track was marked out with cones so that the subject could run a set distance between every transit of the data collection area. This allowed the subject to set a consistent speed with minimal speed variation brought about by stop start running.

It was considered important to control and record running speed during data collection to minimize the effect significant speed differences would have on the stride length and ground reaction force. Photo-electric timer gates (Sportstec) were placed 5m either side of the force plate in the data collection area. These were used to ensure that the subject maintained a constant running speed and had returned to the same speed after a condition change. A trial was not recorded if elapsed time to travel the 10m data collection area was not within 2% of the speed selected by the subject. The elapsed times were recorded for each trial and this was used to calculate average running speed for each trial.

A JVC digital camcorder was used to monitor foot fall during data collection. The camera was directed at right angles to the plane of progression level with the force plate and had a field of view to just below the knee, (See Figure 16 - 19). The video was used to ensure that the foot strikes were made with the heel and that the foot was in contact with the force plate through the whole of the stance phase.

During data collection, trials were rejected if a foot strike was seen to be off the plate, and visual inspection allowed left and right foot strikes to be recorded with trial numbers.

Preparation

The subject was given time to warm up with a light run for 5 minutes, followed by gentle stretching. The procedures for the study were explained to the subject and a number of practice runs conducted. A method of running back over the plate and marking a point 2 strides away from the force plate was used to allow the subject to target away from the data collection area. The procedure was repeated for the faster running speed to accommodate the larger stride length. The subject had little trouble striking the plate consistently with either foot using the technique of targeting a marker placed 2 strides before the plate.

Protocol

The protocol in this study was designed to collect GRF data from both the left and right foot while running with and without foot orthoses. Two speed conditions were used. The protocol was as follows:

Four left foot strikes running in shoes only.

Four right foot strikes running in shoes only.

Four left foot strikes running in shoes with orthoses.

Four right foot strikes running in shoes with orthoses.

Four left foot strikes running at a faster pace in shoes only.

Four right foot strikes running at a faster pace in shoes only.

Four left foot strikes running at a faster pace in shoes with orthoses.

Four right foot strikes running at a faster pace in shoes with orthoses.

A short rest (approximately 60s) was taken at each condition change and a longer (approximately 2min) rest was taken before the protocol was repeated. This gave a total of 64 foot strikes. (8 for each foot in each condition) over a total distance of approximately 2.5km.

Procedure

The subject started running at a self-selected training speed and when a consistent speed and good foot strikes were established recording started. The subject was prompted as to which foot to strike the plate with. Ground reaction forces and video data were recorded continuously, and time elapsed between the photo-cells and which foot struck the plate were recorded manually for each trial.

During a short rest after the first condition foot orthoses were placed in each shoe. After any change in condition the subject was prompted back to a consistent speed using the time between gates as a guide. The faster pace condition, was again at a self selected pace and was closer to competition speed. However, space availability in the hall limited the increase in speed the runner could achieve without discomfort.

When analyzing the ground reaction force data the following temporal and force parameters were used.

Vertical ground reaction force (Fz)

T1 – Time from foot contact to peak impact.

P1 – Magnitude of peak impact.

T2 – Time from foot contact to peak vertical force.

P2 – Magnitude of peak vertical force.

Pap0 – Magnitude of vertical force as anteroposterior force curve crosses zero.

See Figure 13.

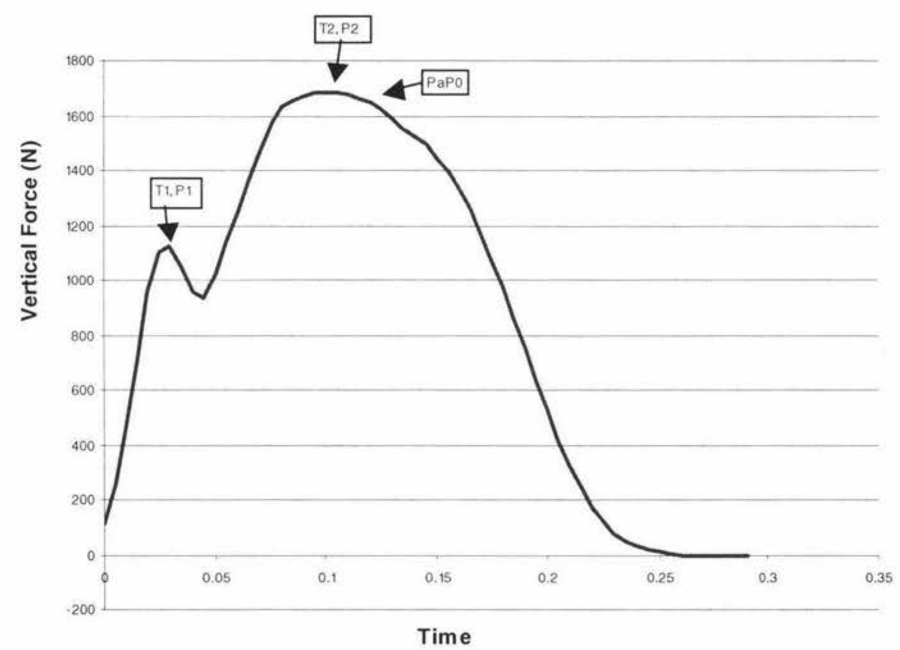


Figure 13 Vertical Ground Reaction Force

Anteroposterior ground reaction force (Fy)

T1 – Time from foot contact to peak braking force.

P1 – Magnitude of peak braking force.

T0 – Time from foot contact to the start of the propulsion phase, when the anteroposterior force curve crosses zero.

see Figure 14.

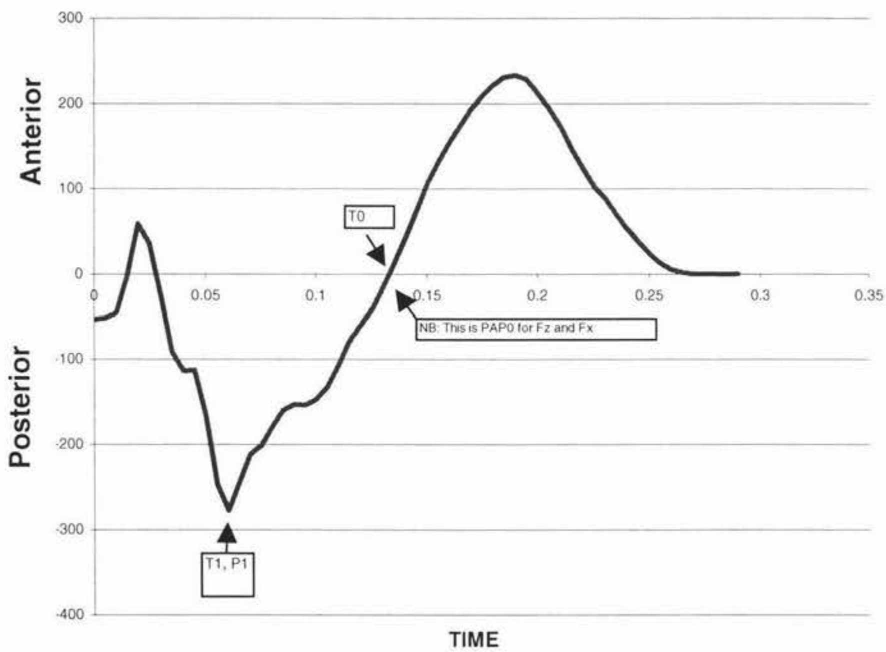


Figure 14 Anteroposterior Ground Reaction Force

Mediolateral ground reaction force (Fx)

T1 – Time from foot contact to first medial peak

P1 – Magnitude of first medial peak

T2 – Time from foot contact to first lateral peak

P2 – Magnitude of first lateral peak

Pap0 – Magnitude of lateral force at the end of the absorption phase, when the anteroposterior force curve crosses zero.

see Figure 15.

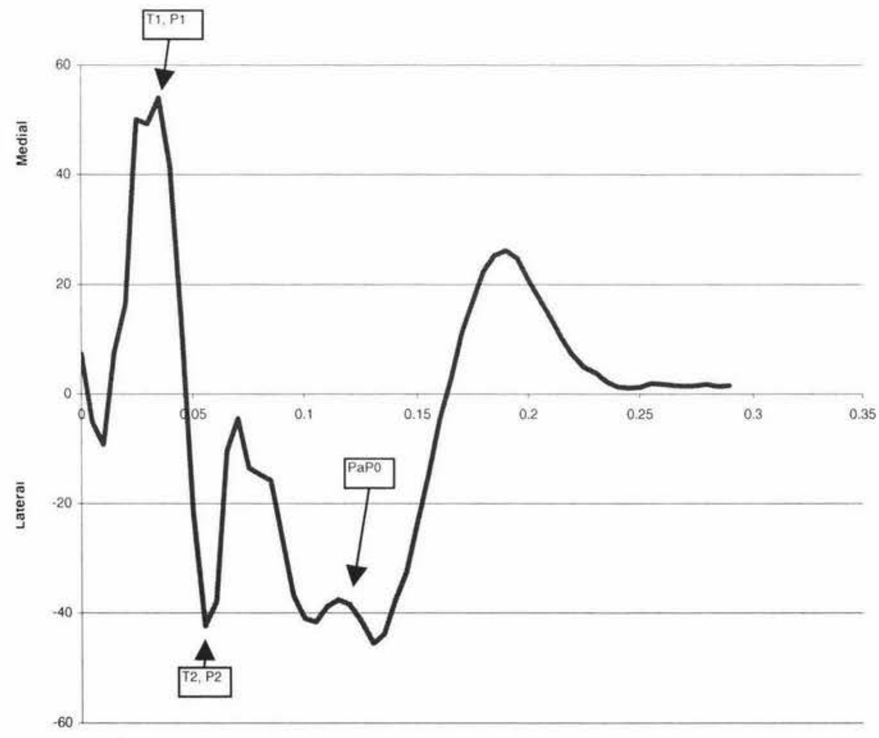


Figure 15 Mediolateral Ground Reaction Force

The photographs in figures 16 to 19 are taken from video footage of the subject during a typical running step.

Figure 16 shows the point of initial contact and the typical heel contact with calcaneal inversion can be seen.

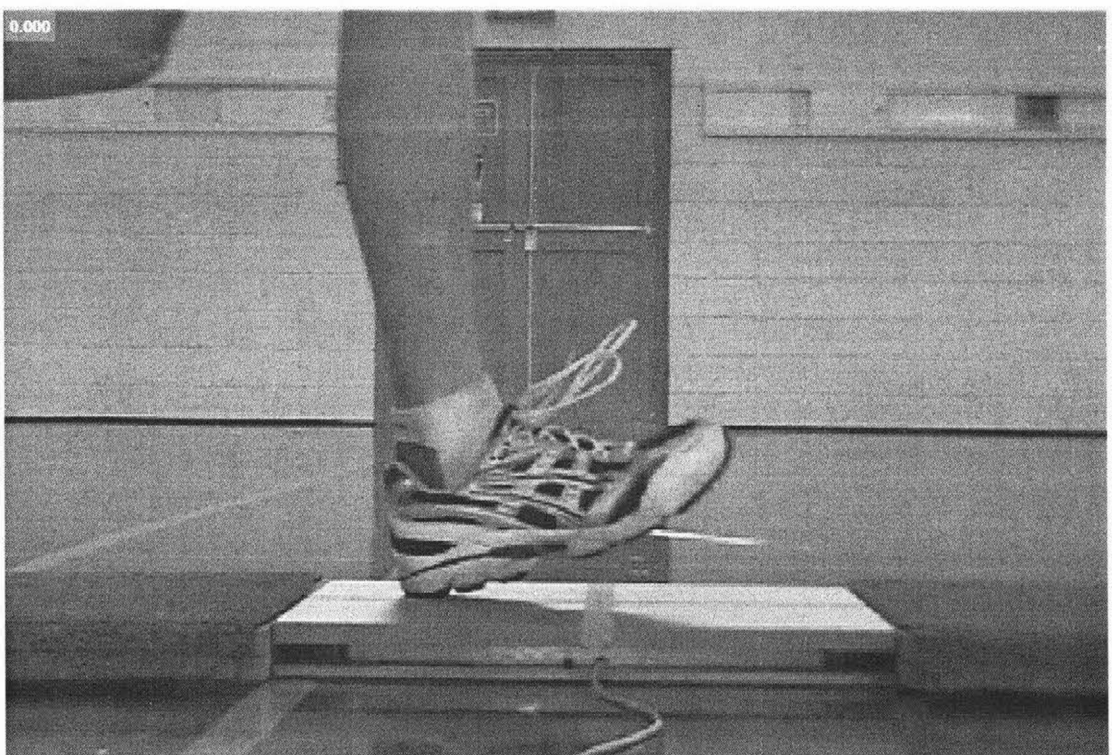


Figure 16 Foot position at heel strike

Figure 17 is 0.020s after heel contact and is approximately the position of T1 (the initial impact peak) for the vertical and mediolateral ground reaction forces.

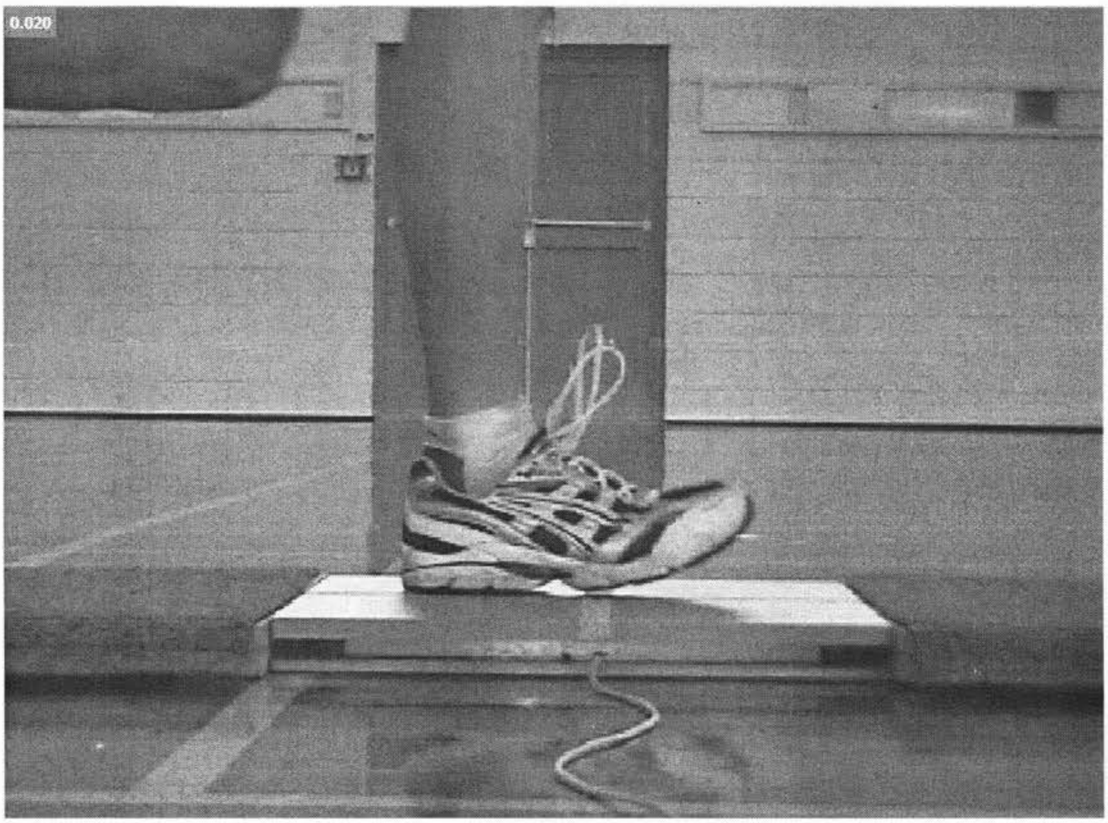


Figure 17 Approximate foot position at T1 for Vertical and Mediolateral Ground Reaction Forces

Figure 18 is 0.060s after heel contact and is approximately the position of T1 (the peak posterior force) for anteroposterior ground reaction force. The foot is seen to be flat on the support surface.

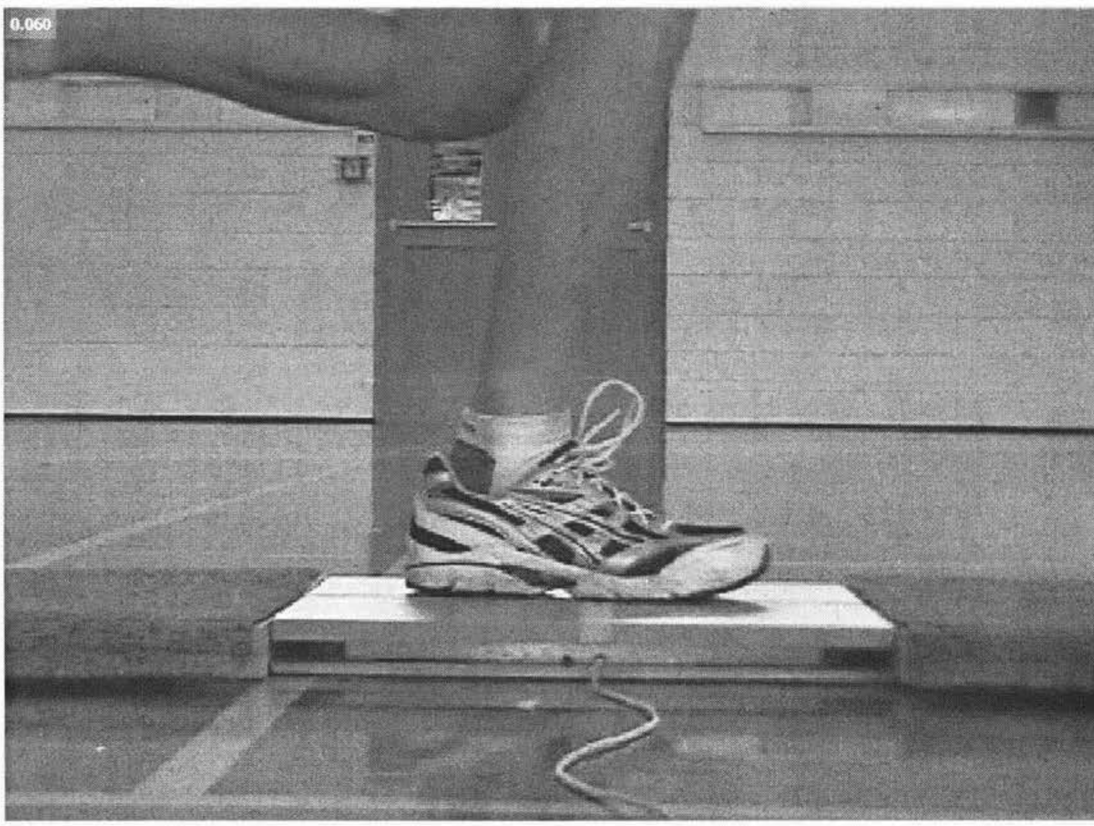


Figure 18 Approximate foot position at T1 Anteroposterior Ground Reaction Force

Figure 19 is 0.140s after heel strike and is approximately the position of T0 for anteroposterior and PAP0 for the vertical and mediolateral ground reaction forces. The heel can be seen to be lifted from the support surface.

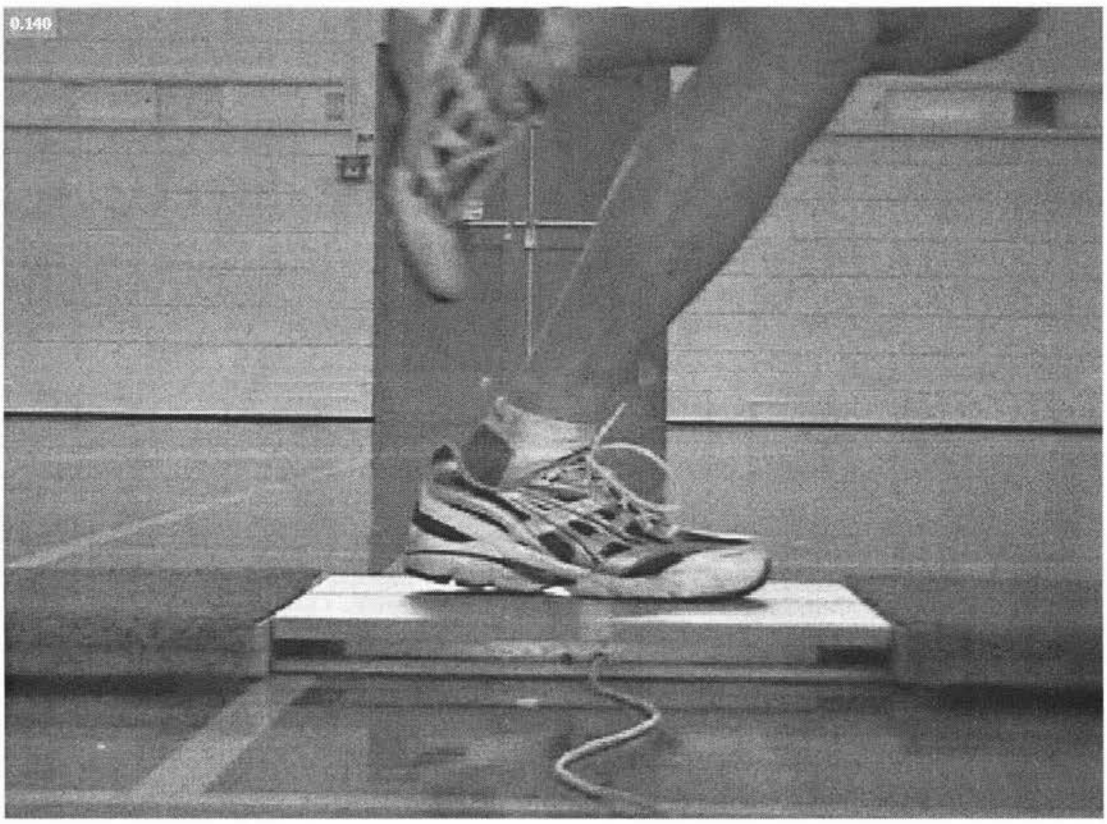


Figure 19 Approximate foot position at T0 for Anteroposterior and PAP0 for Vertical and Mediolateral Ground Reaction Forces

Results

Vertical, anteroposterior and mediolateral ground reaction forces were recorded during running. Past studies have focused mainly on the vertical and, to a lesser degree, the anteroposterior components of ground reaction force because these components are fairly stable and do not vary a large amount between trials. For this study the mediolateral component of the ground reaction force was the main focus of analysis to gain some insight into the mediolateral shear forces at the subtalar joint.

The current study was only concerned with the absorption period of the stance phase of running gait because it was felt that, if foot orthoses were to alter ground reaction force, changes would be seen in the absorption phase of stance. Once the stance phase of gait becomes propulsive the foot would be expected to supinate and the heel will lift off the ground. Once the rear foot clears the ground the orthosis will no longer be able to control the subtalar joint.

The data for left and right foot were combined to give a total of 32 trials for the shod condition and 32 trials for the orthotic condition. The data were then analyzed using SPSS statistical software with a two way anova being performed to assess the effect of orthoses and running speed on ground reaction forces.

Analysis of Vertical Ground Reaction Force

The absorption phase of the vertical ground reaction force was characterized by an impact peak that occurred 0.029s after foot strike for the slow shod condition and 0.028s after foot strike for fast shod condition. The peak impact force was 1107N in the slow shod condition and increased to 1253N in the fast shod condition.

Following the impact spike, the maximum vertical force occurred at 0.103s for the slow shod condition and 0.088s in the fast shod condition. This point in the vertical reaction force was reached in the absorption phase of stance and measured 1727N for the slow shod condition and 1745N for the fast shod condition. The vertical force then declined and at the time of absorption / propulsion cross over it measured between 1600N for the slow shod condition and 1606N for the fast shod condition, (See Figure 20 & Table 1).

These findings were well supported by previous studies looking at the Fz component of ground reaction forces. (Mundermannd et al, 2003)

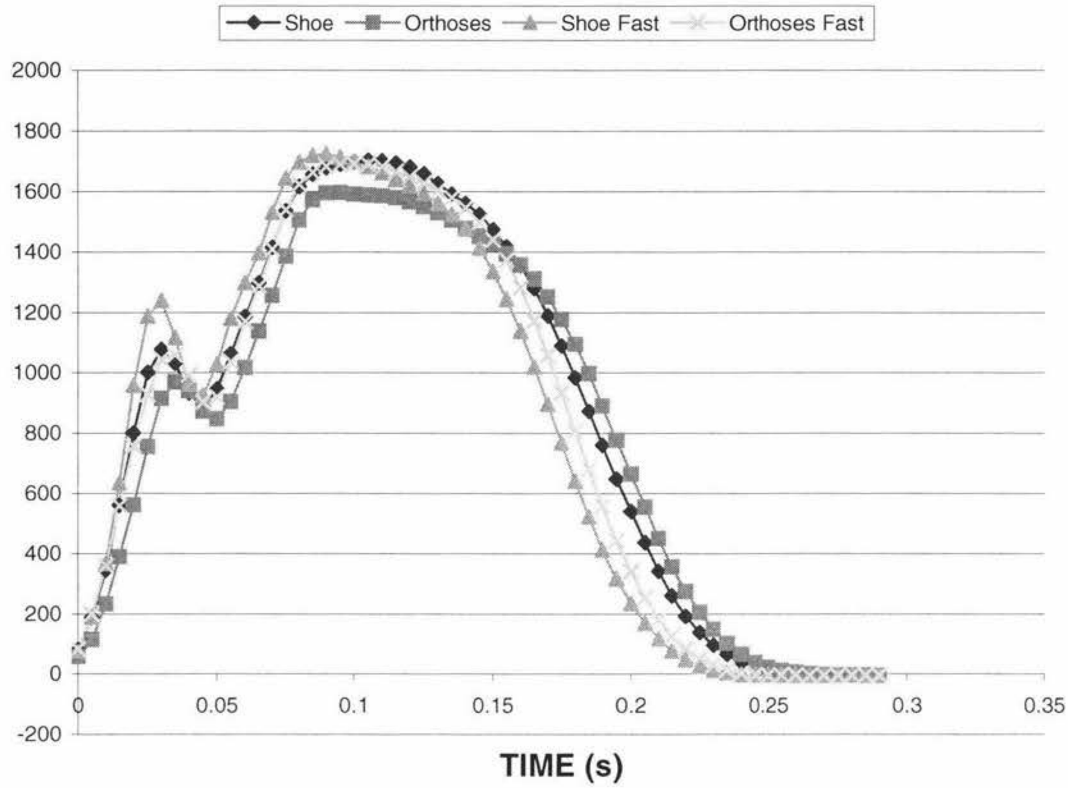


Figure 20 Mean Vertical Ground Reaction Force

Table 1 Descriptive Statistics for Vertical Ground Reaction Force

	Condition	Slow/Fast	Mean	Std. Deviation	N
T1	Shod	Slow	.02951	.004315	16
		Fast	.02875	.002887	16
		Total	.02913	.003632	32
	Orthoses	Slow	.03656	.005072	16
		Fast	.03344	.004366	16
		Total	.03500	.004919	32
P1	Shod	Slow	1107.169	132.40797	16
		Fast	1253.470	98.11530	16
		Total	1180.320	136.61929	32
	Orthoses	Slow	985.0900	121.42095	16
		Fast	1079.510	136.81240	16
		Total	1032.300	135.98295	32
T2	Shod	Slow	.1035	.01397	16
		Fast	.0884	.00908	16
		Total	.0960	.01390	32
	Orthoses	Slow	.0959	.01294	16
		Fast	.0975	.01342	16
		Total	.0967	.01299	32
P2	Shod	Slow	1727.026	107.11966	16
		Fast	1745.934	143.29060	16
		Total	1736.480	124.81758	32
	Orthoses	Slow	1623.530	91.82527	16
		Fast	1728.560	118.89212	16
		Total	1676.045	117.33019	32
PaP0	Shod	Slow	1600.927	144.84170	16
		Fast	1606.365	174.75436	16
		Total	1603.646	157.91072	32
	Orthoses	Slow	1464.288	125.66213	16
		Fast	1607.603	182.06922	16

Table 2 Difference between Shod and Orthotic Condition

Orthotic - Shod			Dependent Variable				
			T1	P1	T2	P2	PaP0
Level 2 vs. Level 1	Contrast Estimate		.006	-148.02	.001	-60.43	-67.70
	Hypothesized Value		0	0	0	0	0
	Difference (Estimate - Hypothesized)		.006	-148.02	.001	-60.43	-67.70
	Std. Error		.001	30.776	.003	29.201	39.619
	Sig.		.000	.000	.814	.043	.093
	95% Confidence Interval for Difference	Lower Bound	.004	-209.58	-.006	-118.84	-146.95
		Upper Bound	.008	-86.45	.007	-2.02	11.55

Table 3 Difference between Fast and Slow Running Speed

Fast - Slow			Dependent Variable				
			T1	P1	T2	P2	PaP0
Level 2 vs. Level 1	Contrast Estimate		-.002	120.36	-.007	61.96	74.37
	Hypothesized Value		0	0	0	0	0
	Difference (Estimate - Hypothesized)		-.002	120.36	-.007	61.96	74.37
	Std. Error		.001	30.776	.003	29.20	39.61
	Sig.		.072	.000	.034	.038	.065
	95% Confidence Interval for Difference	Lower Bound	-.004	58.79	-.013	3.55	-4.87
		Upper Bound	.000	181.92	-.001	120.38	153.62

Effect of Orthoses

Statistical analysis of the time to impact peak (T1) showed a significant difference between the shod and orthotic condition. The time to impact peak was increased from 0.029s after foot strike to 0.035s with the use of foot orthoses. The time to peak impact was decreased by 0.006s (P=.000), (See tables 1 & 2).

Foot orthoses significantly increased the time to impact peak during both running speeds, which suggests that orthoses are increasing the time over which the foot/shoe unit is absorbing the vertical impact energy.

The magnitude of the impact peak (P1) decreased significantly from 1180N in the shod condition to 1032N in the orthotic condition. The mean total vertical impact force was reduced by 148N (P=.000) when running with orthoses, (See tables 1 & 2).

Maximum vertical ground reaction force occurred before the crossover to the propulsive phase of stance. By this time a significant amount of subtalar joint pronation may have occurred so the magnitude of vertical ground reaction force at this point might indicate the shock absorption effect of the foot orthoses at the end of the pronation phase of stance.

The time from foot contact to maximum vertical ground reaction force (T2) was not significantly altered by the use of foot orthoses. However, the peak vertical force (P2) was significantly reduced with orthotic use. P2 was reduced from 1736N to 1676N by the use of foot orthoses. The mean total vertical force was reduced by 60N (P=.043) by orthotic use, (See tables 1 & 2).

The fact that the maximum vertical ground reaction force was significantly reduced by orthoses indicates that energy transfer following the initial impact was modified by the use of orthoses. This may mean that in this subject the foot orthoses have improved the shock attenuating function of the feet during the end phase of foot pronation while running.

The last point in the vertical ground reaction force curve that was analyzed was the magnitude of the force at the time of the end of the absorption phase of gait (PaP0). PaP0 was indicated by the time that the anteroposterior ground reaction force curve crossed zero. This point was in the declining part of the vertical force curve. No significant changes were seen in PaP0 with the use of foot orthoses. These results indicate that at the time the foot becomes propulsive the vertical force is not altered by orthoses use, (See tables 1 & 2).

When looking at these results it appears that the use of orthoses has some effect on the foot in the initial impact peak and the absorption phase of gait.

Effect of Speed

The average slow running speed across both conditions was 3.78m.s with a SD of 0.06m.s (CV 1.6%), while the faster running speed across both conditions was 4.33m.s with a SD of 0.04m.s (CV 0.9%).

The effect of speed on the time to peak impact vertical force (T1) showed no significant change, however the peak impact force (P1) was significantly changed by running speed. In the shod condition the faster running speed increased the vertical impact peak from 1107N to 1253N while faster running during the orthotic condition increased P1 from 985N to 1079N. The faster running speed showed an increased impact peak of 120N (P=.000).

The time to the maximum vertical force (T2) was reduced with faster running speed by 0.007s (P=.034). Faster running reduced T2 from 0.103s to 0.088s in the shod condition but increased it from 0.095s to 0.097s in the orthotic condition. The peak vertical force (P2) was increased by 61N (P=.038) while running at the faster speed. During the shod condition faster running increased P2 from 1727N to 1745N while in the orthotic condition faster running increased P2 from 1623N to 1728N. The vertical force at PaP0 increased with the faster speed but did not reach significance (P=0.065), (See table 3).

These results are to be expected because as running speed increases stance time reduces and vertical ground reaction forces would be expected to increase.

Interaction between Condition and Speed

Table 4 Interaction between Condition and Speed

Condition X Speed	Variable	F	Sig
	T1	1.23	.268
	P1	0.71	.403
	T2	7.09	.010
	P2	2.17	.146
	PaP0	3.02	.087

The only significant interaction between condition and speed was seen in the time to the maximum vertical force (T2) (P=.010).

The orthotic condition showed no increase in the time to peak vertical force while running with orthoses. The time to peak vertical force in the faster condition reduced significantly by 0.007s ($P=.034$), (See table 4)

Analysis of the Anteroposterior Ground Reaction Force.

The absorption phase of the anteroposterior ground reaction force was characterized by a rise in posterior force to a peak at 0.058s after foot contact in the slow shod condition and 0.057s in the fast shod condition. The magnitude of the peak posterior force was -296N in the slow shod condition and –361N in the fast shod condition.

The force curve reversed and crossed zero at 0.116s after foot strike in the slow shod condition and 0.122s after foot strike in the fast shod condition.

The point at which the anteroposterior force became anterior marked the end of the absorption phase of stance,(see Figure 21 & Appendix)

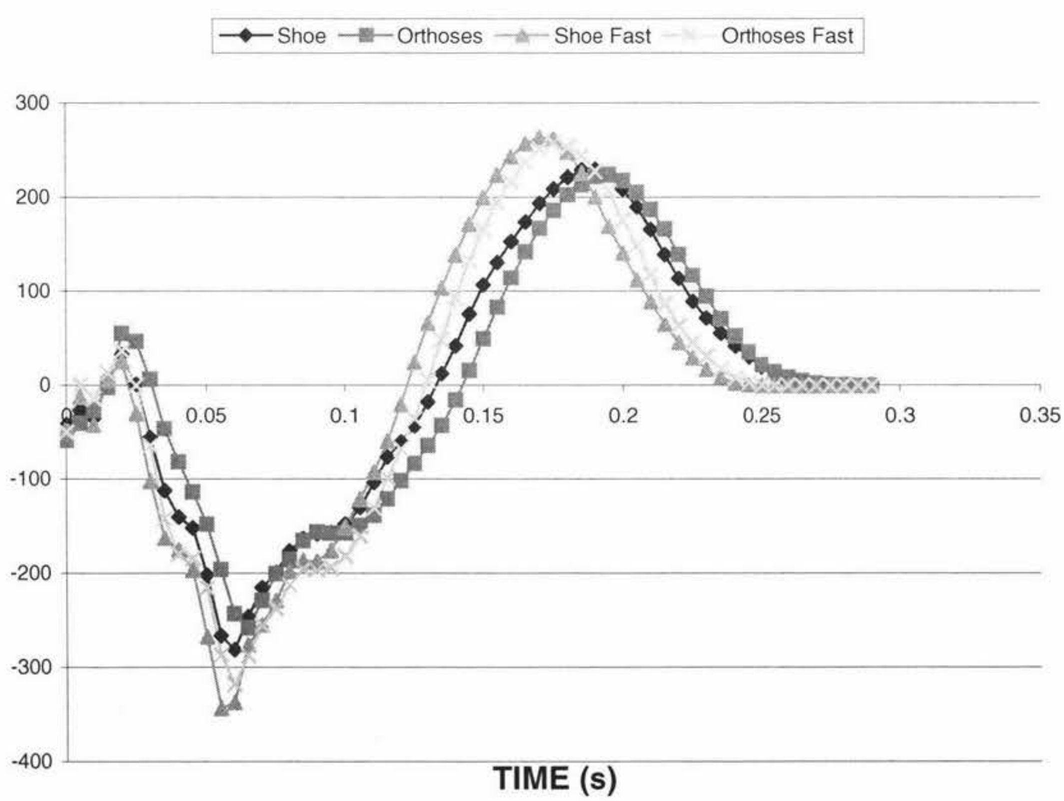


Figure 21 Mean Anteroposterior Ground Reaction Force

Table 5 Descriptive Statistics for Anteropostior Ground Reaction Force

	Condition	Slow/Fast	Mean	Std. Deviation	N
T1	Shod	Slow	.05866	.003399	16
		Fast	.05719	.003637	16
		Total	.05792	.003543	32
	Orthoses	Slow	.06469	.002213	16
		Fast	.06063	.003096	16
		Total	.06266	.003356	32
P1	Shod	Slow	-296.1805	22.71591	16
		Fast	-361.4100	20.68111	16
		Total	-328.7953	39.42935	32
	Orthoses	Slow	-266.2725	24.11636	16
		Fast	-329.1019	18.29897	16
		Total	-297.6872	38.23822	32
T0	Shod	Slow	.1168	.06725	16
		Fast	.1225	.00577	16
		Total	.1196	.04704	32
	Orthoses	Slow	.1422	.00752	16
		Fast	.1291	.00491	16
		Total	.1356	.00914	32

Table 6 Differences between Shod and Orthotic Condition

Orthotic - Shod			Dependent Variable		
			T1	P1	T0
Level 2 vs. Level 1	Contrast Estimate		.005	31.108	.016
	Hypothesized Value		0	0	0
	Difference (Estimate - Hypothesized)		.005	31.108	.016
	Std. Error		.001	5.391	.009
	Sig.		.000	.000	.065
	95% Confidence Interval for Difference	Lower Bound	.003	20.324	-.001
		Upper Bound	.006	41.892	.033

Table 7 Differences between Fast and Slow running Speed

Fast - Slow			Dependent Variable		
			T1	P1	TO
Level 2 vs. Level 1	Contrast Estimate		-.003	-64.029	-.004
	Hypothesized Value		0	0	0
	Difference (Estimate - Hypothesized)		-.003	-64.029	-.004
	Std. Error		.001	5.391	.009
	Sig.		.001	.000	.665
	95% Confidence Interval for Difference	Lower Bound	-.004	-74.813	-.021
		Upper Bound	-.001	-53.245	0.13

Effect of Orthoses

The time to maximum posterior force (T1) was significantly altered by the use of foot orthoses. T1 was increased from 0.057s after foot contact to 0.062s after foot contact with the use of foot orthoses. The time to the peak posterior force was increased by 0.005s (P=0.000), (See table 5 & 6).

The peak posterior force (P1) was significantly decreased by the use of foot orthoses. P1 was reduced from –328N in the shod condition to –297N in the orthotic condition. These data show a decrease in maximum posterior force of 31N (P=0.000), (See tables 5 & 6).

The time at which the anteroposterior ground reaction force crossed zero (T0) showed no change with the use of foot orthoses.

Effect of Speed

Analysis of the effect of speed on the peak posterior force showed significant changes in the time to peak (T1). In the shod condition the faster running speed decreased the time to T1 from 0.058s to 0.057s while in the orthotic condition the faster running speed decreased the time to peak posterior force from 0.064s to 0.060s (P=0.001), (See tables 5 & 7).

The peak posterior force (P1) was significantly increased by the faster running speed. In the shod condition the faster running speed increased the posterior peak from –296N to –361N while in the orthotic condition the faster running speed increased the posterior peak from –266N to –329N. The faster running speed showed an increase in the peak posterior forces of 64N (P=0.000), (See tables 5 & 7).

The time at which the anteroposterior ground reaction force crossed zero was not changed by increasing running speed from 3.78m.s-1 to 4.35m.s-1.

Interactions between Condition and Speed

Table 8 Interaction between Condition and Speed

Condition X Speed	Variable	F	Sig
	T1	2.732	.104
	P1	.050	.825
	TO	1.225	.273

No significant interactions were seen between the orthotic conditions and speed in the anteroposterior ground reaction force, (See Table 8).

Analysis of the Mediolateral Ground Reaction Force

The absorption phase of the mediolateral ground reaction force was characterized by medial impact peak at 0.027s after foot strike in the slow shod condition and 0.024s after foot strike in the fast condition. The magnitude of this force was 70N in the slow shod condition and 83N in the fast shod condition.

The force curve rapidly reversed to form a lateral peak with a magnitude of 60N in the slow shod condition and 57N in the fast shod condition. This peak was seen at 0.055s after heel strike in both the slow and fast shod conditions. The force curve reduced slightly after this peak and then returned to a similar magnitude, which was maintained until the end of the absorption phase.

At the time of cross over from absorption to propulsion the magnitude of the lateral force was 45N in the slow shod condition and 39N in the fast shod condition. The mediolateral force curve reversed to medial during the propulsive phase of stance.

Previous work on mediolateral ground reaction force has suggested that the force is highly subject specific but the general trend is consistent with the literature. (Hamill & Knutzen (1995), (see Figure 22 & 23 & Appendix 24 & 25)

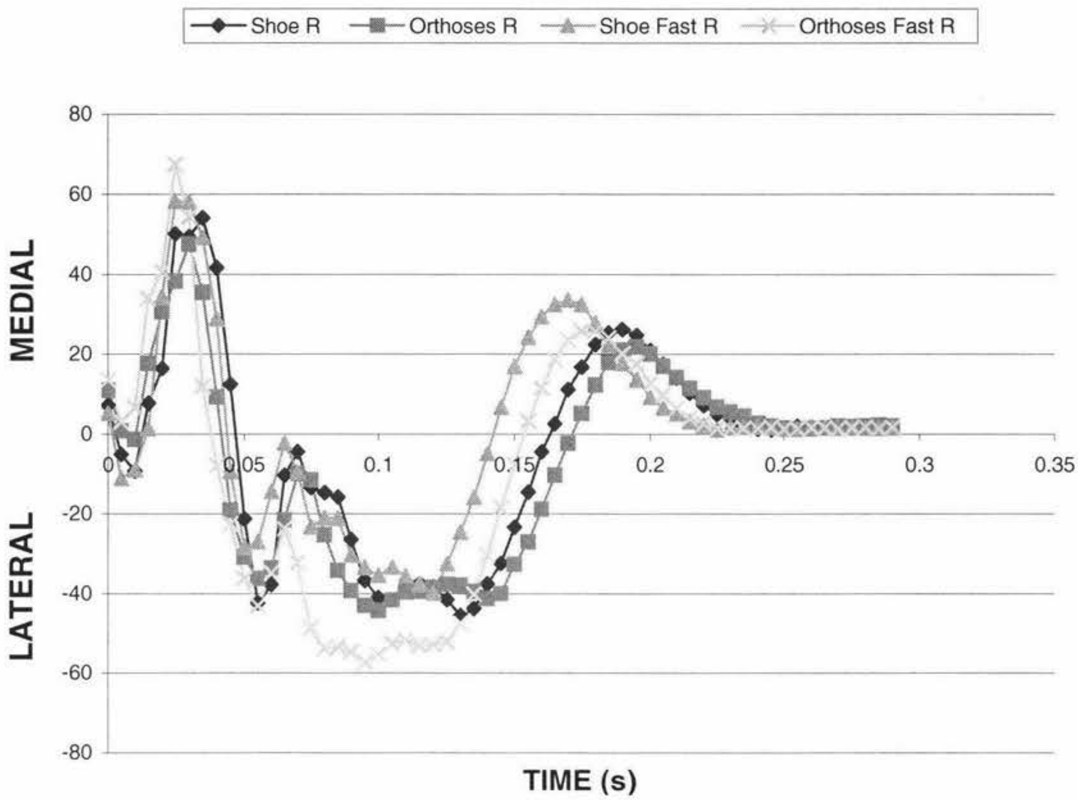


Figure 22 Mean Mediolateral Ground Reaction Force for Right Foot

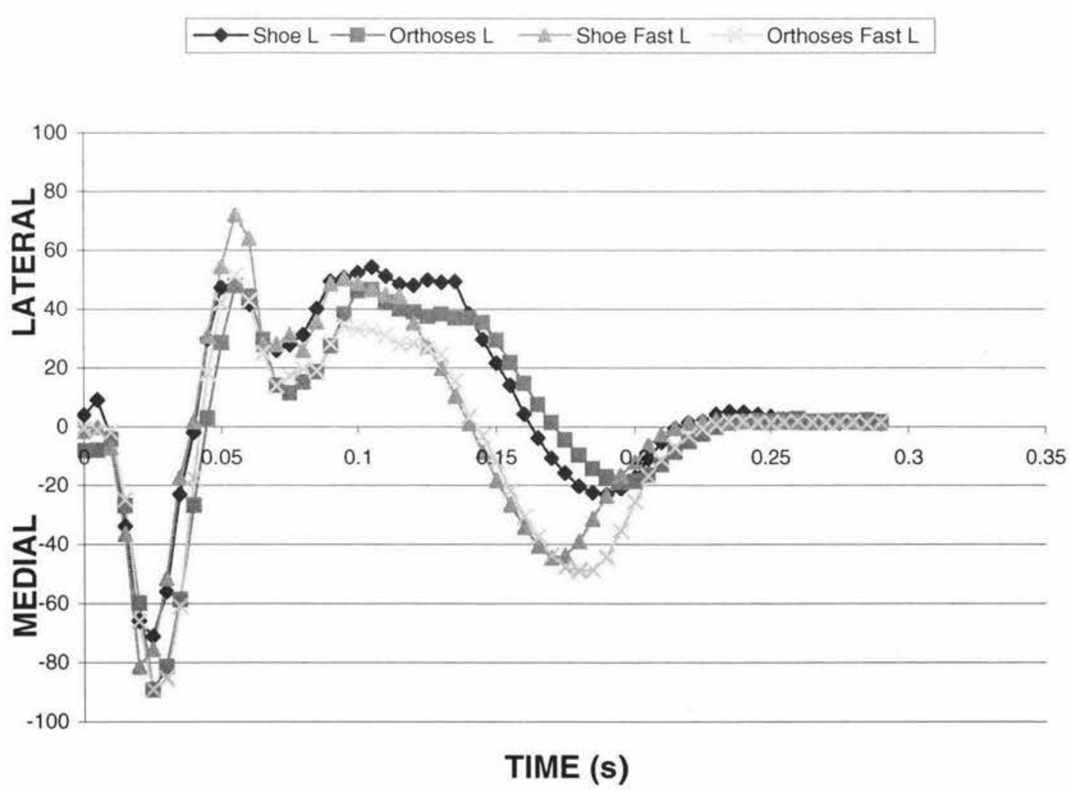


Figure 23 Mean Mediolateral Ground Reaction Force for Left Foot

Table 9 Descriptive Statistics for Mediolateral Ground Reaction Force

	Condition	Slow/Fast	Mean	Std. Deviation	N
T1	Shod	Slow	.02741	.005826	16
		Fast	.02438	.004425	16
		Total	.02589	.005317	32
	Orthoses	Slow	.02750	.004830	16
		Fast	.02563	.002500	16
		Total	.02656	.003902	32
P1	Shod	Slow	70.2274	16.46961	16
		Fast	83.3931	23.49704	16
		Total	76.8103	21.05068	32
	Orthoses	Slow	70.6838	27.28655	16
		Fast	79.1250	19.43302	16
		Total	74.9044	23.69361	32
T2	Shod	Slow	.05549	.006059	16
		Fast	.05562	.004031	16
		Total	.05556	.005063	32
	Orthoses	Slow	.05594	.004171	16
		Fast	.05250	.008756	16
		Total	.05422	.006969	32
P2	Shod	Slow	60.1114	31.90846	16
		Fast	57.6844	30.00957	16
		Total	58.8979	30.49484	32
	Orthoses	Slow	47.0688	19.19782	16
		Fast	48.4462	18.44833	16
		Total	47.7575	18.53387	32
PaP0	Shod	Slow	45.6200	28.67590	16
		Fast	39.4325	20.56530	16
		Total	42.5263	24.74702	32
	Orthoses	Slow	42.2681	20.62095	16
		Fast	39.2313	21.54001	16
		Total	40.7497	20.79990	32

Table 10 Differences between Shod and Orthotic Condition

Orthotic - Shod			Dependent Variable				
			T1	P 1	T2	P2	PaP0
Level 2 vs. Level 1	Contrast Estimate		.001	-1.906	-.001	-11.140	-1.777
	Hypothesized Value		0	0	0	0	0
	Difference (Estimate - Hypothesized)		.001	-1.906	-.001	-11.140	-1.777
	Std. Error		.001	5.514	.002	6.408	5.775
	Sig.		.559	.731	.380	.087	.759
	95% Confidence Interval for Difference	Lower Bound	-.002	-12.935	-.004	-23.957	-13.328
		Upper Bound	.003	9.123	.002	1.677	9.775

Table 11 Differences between Fast and Slow Running Speed

Fast - Slow			Dependent Variable				
			T1	P1	T2	P2	PaP0
Level 2 vs. Level 1	Contrast Estimate		-.002	10.803	-.002	-.525	4.612
	Hypothesized Value		0	0	0	0	0
	Difference (Estimate - Hypothesized)		-.002	10.803	-.002	-.525	4.612
	Std. Error		.001	5.514	.002	6.408	5.775
	Sig.		.035	.055	.280	.935	.428
	95% Confidence Interval for Difference	Lower Bound	-.005	-.225	-.005	-13.342	-16.164
		Upper Bound	.000	21.832	.001	12.292	6.940

Effect of Orthoses

No significant changes were seen in the mediolateral ground reaction force with the use of foot orthoses. This suggests that the shear force that act at almost 90 degrees to the subtalar joint in the frontal plane have not been altered by foot orthoses during the time of rapid pronation of the foot. (See tables 9 & 10).

Effect of Speed

Analysis of time to peak medial shear (T1) was significantly altered by an increase in running speed. In the shod condition the faster running speed decreased the time to T1 from 0.029s to 0.024s while in the orthotic condition the faster running speed reduced T1 from 0.027s to 0.025s. This means that faster running speed decreased the time to peak medial shear by 0.002s (P=0.023), (See table 11).

Interaction between Condition and Speed

Table 12 Interaction between Condition and Speed

Condition X Speed	Variable	F	Sig
	T1	.259	.612
	P1	.184	.670
	T2	1.388	.243
	P2	.088	.768
	PaP0	0.74	.786

No significant interactions were seen between the orthotic condition and speed in the mediolateral ground reaction force, (See Table 12).

Running Speed

Running speed was recorded to ensure that changes seen in ground reaction forces were not just a result of speed fluctuations, (see Table 13).

Table 13: Average Running Speed Recorded in Blocks of 8 Trials

Condition	Trial	Mean Speed ($m.s^{-1}$)	Mean of Trials	SD	CV
Shod Slow	Trials 1 - 8	3.73			
Shod Slow	Trials 9 - 16	3.85	3.79	0.07	1.8%
Orthoses Slow	Trials 1 - 8	3.75			
Orthoses Slow	Trials 9 - 16	3.78	3.76	0.02	0.5%
Shod Fast	Trials 1 – 8	4.37			
Shod Fast	Trials 9 - 16	4.31	4.34	0.04	0.9%
Orthoses Fast	Trials 1 – 8	4.37			
Orthoses Fast	Trials 9 - 16	4.29	4.33	0.06	1.4%

The average slow running speed across both conditions was $3.78m.s^{-1}$ with a SD of $0.06m.s^{-1}$ (CV 1.6%), while the faster pace across both conditions was $4.33m.s^{-1}$ with a SD of $0.04m.s^{-1}$ (CV 0.9%). This degree of running speed consistency compared well with the literature and so it was considered that the effect of variation in running speed within a condition would be minimal.

Discussion

This study was conducted to examine the possible acute changes in ground reaction forces during the absorption phase of running gait induced by wearing foot orthoses. The design used was a case study with the subject acting as his own control. This design was used to counter the effect of the high inter-subject variability observed in ground reaction forces, in particular the mediolateral component. In spite of the single subject design the results compared well with previous studies, which included more subjects, (Mundermann et al, 2003; Morarty & Agosta, 1998).

The time to peak vertical impact force (T1) was significantly increased while the magnitude of the vertical impact peak (P1) was significantly decreased with the use of foot orthoses, which compared well with literature (Mundermann et al, 2003). The first peak seen in the mediolateral ground reaction force (T1) was in a medial direction and occurred at a similar time to that of the impact peak of the vertical ground reaction force, which suggests that it was a consequence of the initial impact of the foot. Unlike the vertical impact peak, the first mediolateral peak did not differ in the time to peak (T1) or the peak magnitude of the force (P1) with the use of foot orthoses even though changes at this phase of stance might have been expected because the orthosis had a 4^0 medial rear foot wedge, which would have placed the calcaneus in a more inverted position.

These results suggest that, during the initial impact, the orthoses slowed the rate of energy transfer between the leg, the shoe/foot unit, and the ground, but did not alter the shear forces acting approximately at right angles to the subtalar joint.

This means that something in the initial impact chain had absorbed shock. The mechanism for this increased shock absorption could be;

Mechanical – changes in joint function;

Compliance – either soft tissue or orthotic; or

Neural – proprioceptive changes to foot and leg function.

Novacheck (1998), in a review paper, suggested that pronation unlocks the mid-tarsal joint, increasing the flexibility of the foot, which allows it to function as an effective shock absorber. This would suggest the subtalar joint is important in transmission of ground reaction forces during the stance phase of gait, and so it could mean that the control of pronation might be the mechanism by which foot orthoses enhance impulse control and shock absorption.

However, Henning & Milani (1998), commenting on Edington, Frederick & Cavangh (1990), concluded that in heel toe running the rear foot is typically dorsiflexed and inverted at initial contact. The foot then starts to plantarflex and pronate, but at approximately 30ms after heel strike significant rear foot inversion still exists. This means that, at the time of the passive vertical impact peak (27ms to 39 ms after first contact in this study), little pronation would have occurred. Because only a small amount of total pronation occurs during the passive impact phase of stance, it was unlikely to be subtalar joint motion that was responsible for the reduction in initial loading rate seen in this study.

Mundermann et al (2003), when studying the effect of medial foot wedges, molded orthotics and posted orthotics saw decreases in loading rate and peak vertical impact force in the molded and posted orthotic conditions. However, they saw no changes in rear foot eversion, which suggests decreasing pronation was not the cause of the effect. When Mundermann et al (2003) used medial wedges with no orthotic construction they saw a significant reduction in rear foot eversion but, in contrast to the orthotic and posted orthotic condition, they saw increased loading rates and an increase in maximum vertical impact force. This finding suggests that, by decreasing rear foot eversion during the passive impact phase of absorption, shock absorption may be reduced. Similar results were seen by Perry & Lafortune (1995), when studying the effects of inverted, neutral and everted shoes during running. They also found that the inverted shoe that significantly reduced rear foot eversion also increased impact loading. This literature suggests that the changes in impact loading reported here were not a result of reduced pronation because, with reduced rear foot eversion increased impact loading would have been expected.

The second possible cause of the reduced shock loading could be changed compliance during the impact phase of stance. The two structures that might have produced changes in compliance would be the orthotic and the heel fat pad.

The foot orthoses used in this study had no specific shock absorbing characteristics but with the EVA wedge they would still be expected to change compliance to a small degree. However Mundermann et al (2003), put structures under the heel in both the wedging and orthotic conditions and only saw a decrease in impact loading in the orthotic condition. Consequently, it appears that increased compliance, created by additional material under the heel, did not reduce the impact loading because the reduction would have been seen in both conditions.

In a review article Whittle (1999), commented that wearing shoes contained the heel fat pad, which prevented lateral deformation and reduced the amount that it compressed when compared with bare feet walking. Winter et al (1996), considered that a large amount of energy is absorbed in the fat pad of the heel with much less being absorbed in the inverter and dorsiflexor muscles of the foot. Consequently, cupping of the heel by the orthosis could produce a less compliant heel by preventing lateral deformation and complete compression of the fat pad.

As the foot progresses through the absorption phase of gait the anteroposterior ground reaction force reaches its posterior peak. The subject in this study had a significant increase in the time to maximum posterior force (T1) and a decrease in the magnitude of the posterior force (P1) with the use of foot orthoses. At a similar time to the peak posterior force the second peak in the mediolateral ground reaction force (T2) was seen and was in a lateral direction. As the mediolateral force was in a lateral direction its effect on the subtalar joint would be to pronate it. No significant changes were seen in the time to peak or the magnitude of the lateral peak of the mediolateral force with the use of orthoses. The peak vertical ground reaction force (P2) was reached after the peak posterior and lateral shear forces. The time to peak vertical force (T1) was not changed with the use of foot orthoses, which was consistent with current studies, (Mundermann et al, 2003). However, the magnitude of the peak vertical force was significantly reduced with the use of foot orthoses, which was not seen by Mundermann et al (2003). This reduced vertical force suggests that energy is being absorbed either in a similar way to the passive impact phase or by flexion of the semi-flexible orthotic plate. As the flexibility of the orthotic plate is not identical to previous research it would be expected that variations might be seen in the later absorption phase of stance. This variation seen in peak vertical ground reaction force maybe caused by a less compliant orthotic plate than Mundermann et al (2003), used in their study, which could assist the arch in energy transfer.

The phase of gait at which the peak posterior shear, peak lateral shear and peak vertical ground reaction force occur is referred to as active absorption because the muscles of the lower limb are working to moderate net forces and control motion.

Because the active absorption phase of stance is the time at which most of the pronation of the foot seems to occur it would be expected that any changes to ground reaction forces induced by the foot orthoses acting to control pronation would be seen at this time. These results show significant changes only in the vertical and posterior force, which suggests that the foot orthoses did not change the way in which the body accommodates lateral forces while the body is actively involved in deceleration of the lower limb and center of mass. However, the changes in the vertical and posterior force may indicate some change in lower limb kinematics. As in the passive absorption phase, no changes were observed in the mediolateral forces that would have most effect on the subtalar joint. This suggests that frontal plan kinematics of the foot were unchanged during the active absorption phase of stance, which implies that pronation was also unchanged.

Saltzman & Nawoczenski (1995) commented that peak pronation was seen between 35% and 45% of the stance phase of gait, and so it would be expected that, if pronation resulted in reduced total vertical reaction forces, a reduction would be seen in the maximum vertical force. The fact that this study did see a change in vertical ground reaction force during the active absorption phase of stance means that foot orthoses absorbed energy or produced kinematic changes in the lower limb that effectively reduced the rate of energy transfer.

Because Mundermann et al (2003), saw no change in peak vertical force with medial wedging in spite of the fact that it did reduce rear foot eversion this suggests that, even when pronation is reduced, peak vertical forces are not altered, which would indicate that foot function is not able to affect the vertical deceleration of the center of mass.

The last point of the ground reaction force to be analysed was the end of the absorption phase of stance (P0). No significant changes were seen in the time to crossover of the anteroposterior ground reaction force, which shows that the total period of absorption was not changed with the use of foot orthoses. Because the peak posterior force was decreased but the total time to crossover was not changed, it appears that the foot orthoses changed the shape of the force curve but the impulse generated by the posterior force was largely the same. This effect on the anteroposterior force was perhaps related to the changes seen in the impact part of the vertical ground reaction force. It appears that, during the absorption phase of gait, foot orthoses may have reduced the rate at which the impact forces were accepted by the lower limb. Because running speed was consistent between conditions, the change in anteroposterior force parameters could have been caused by changes in lower limb kinematics or muscle activity, which could have resulted in subtle changes to the shape of the anteroposterior ground reaction force but not the duration of the absorption phase.

The vertical and mediolateral ground reaction forces at the point of crossover were not altered by foot orthoses but it was noted that around this point the mediolateral curve was much more variable than in the passive and early active absorption phases. Hamil

& Knutzen (1995), observed that the mediolateral shear was the most variable component of the ground reaction force and so these findings are consistent with the literature, (see Appendix 24 & 25).

The results of this study show that, for this subject, the forces that act in the frontal plane, approximately at right angles to the subtalar joint, were not altered acutely by the use of foot orthoses. It would be expected that, if pronation was reduced by wearing foot orthoses changes would be seen in the shear forces that act in the frontal plane because rear foot eversion is predominantly a frontal plane motion. From the current study, it appears that foot orthoses do not alter the forces that are associated with foot pronation, but may reduce the forces that are related to impact loading. This means that foot orthoses may only have a limited effect on frontal plane motion of the foot during running, which is in broad agreement with recent research looking at foot orthoses, (Belchamber et al, 2000; Mundermann, 2003).

Effect of Running Speed

The effect of foot orthoses on ground reaction forces at two different running speeds was investigated. It was expected that an increase in running speed would increase ground reaction forces and, therefore produce an increase in effect size.

The faster running pace significantly altered the passive impact phase of stance by increasing the magnitude of the vertical impact peak (P2) and decreasing the time to the impact peak of the mediolateral force (T1).

These results would be expected, because at the faster running speed stride length would increase which requires increased ground reaction forces which would increase the impact forces at foot strike. The fact that time to vertical impact peak (T1) was not altered may indicate that the change in vertical loading rate was not detected at a sampling rate of 200 Hz.

In the anteroposterior ground reaction force both the time to peak posterior force (T1) and the magnitude at peak posterior force (P1) were significantly altered by running speed. The faster running pace decreased the time to peak and increased the magnitude of the peak. This result would be expected because the increased speed produced greater impulses in the sagittal plane. The decrease in the time to peak can be explained by a decrease in the duration of stance phase when running at faster speed. The time to the crossover from absorption to propulsion (T0) was decreased by the faster running speed, but this reduction was not significant. We would expect to see a decrease in the time to cross over from absorption to propulsion with a faster running speed because the stance phase of gait is shortened by increased running speed.

During the active absorption phase of stance the time to maximum vertical force (T2) and the magnitude of the maximum impact force (P2) were significantly altered by increased running speed.

The decrease in the time to maximum vertical force is a result of decreased stance time of higher running speeds while the increase in the magnitude of the vertical force was a result of the need to generate a larger vertical impulse to control the vertical speed of the centre of mass.

Conclusion

The results of this research indicate that foot orthoses had no effect on the mediolateral ground reaction force during the absorption phase of running gait. Because the mediolateral ground reaction force acts almost at right angles to the subtalar joint axis this means that the time course or extent of pronation probably was not changed by the use of orthoses. This is, perhaps, not surprising because of the substantial deformability of soft tissue between the calcaneus and the orthotic plate. The pressure concentrated on the medial boarder of the foot will compress the heels fat pad and may limit the effect of the orthotic on the underlying skeletal structures of the foot. This indicates that any clinical benefits seen in the use of foot orthoses may come from changes in structures and functions unrelated to subtalar joint kinematics such as enhanced proprioception and altered muscle recruitment.

Clinically, these findings indicate that, when a podiatrist prescribes orthoses to treat an injury, possibly as a result of training or competition, the foot orthoses would not be expected to limit pronation at the subtalar joint during running. Consequently, if changes are not seen in the ground reaction forces that partly drive the kinematics of the foot, it seems highly unlikely that significant changes will be found in the more proximal joints of the lower limb.

Because clinical benefits from the use of orthoses have been seen in some circumstances it is important to determine the factors that lead to these benefits. This may allow a new style of insole to be designed, which will produce more consistent and reliable results. A possible area for future research might be to investigate the effect of orthotic inserts on muscle activation patterns because foot orthoses may have a proprioceptive effect, or assist in support of the arch, thus reducing load in the muscles that act in arch control.

Understanding how shoe inserts or foot orthoses act to modify foot and lower limb function may provide new tools to assess and treat musculo-skeletal injury of the lower limb. These diagnostic tools may replace the current clinical focus on subtalar joint function. Whatever the possible effects on the kinetics and kinematics of the lower limb it is likely to be highly subject specific given the variation in the subtalar joint axis location (Valmassy,1996).

Because this is a single subject study the findings from this particular subject may not be generalized to the wider population.

A study with a larger sample size may confirm these results in a wider population, but, because of the expected high inter-subject variability seen in the mediolateral ground reaction force, (Hamill & Knutzen, 1995), it is likely that consistent patterns across individuals will not be observed in the magnitude or time course of the mediolateral ground reaction force.

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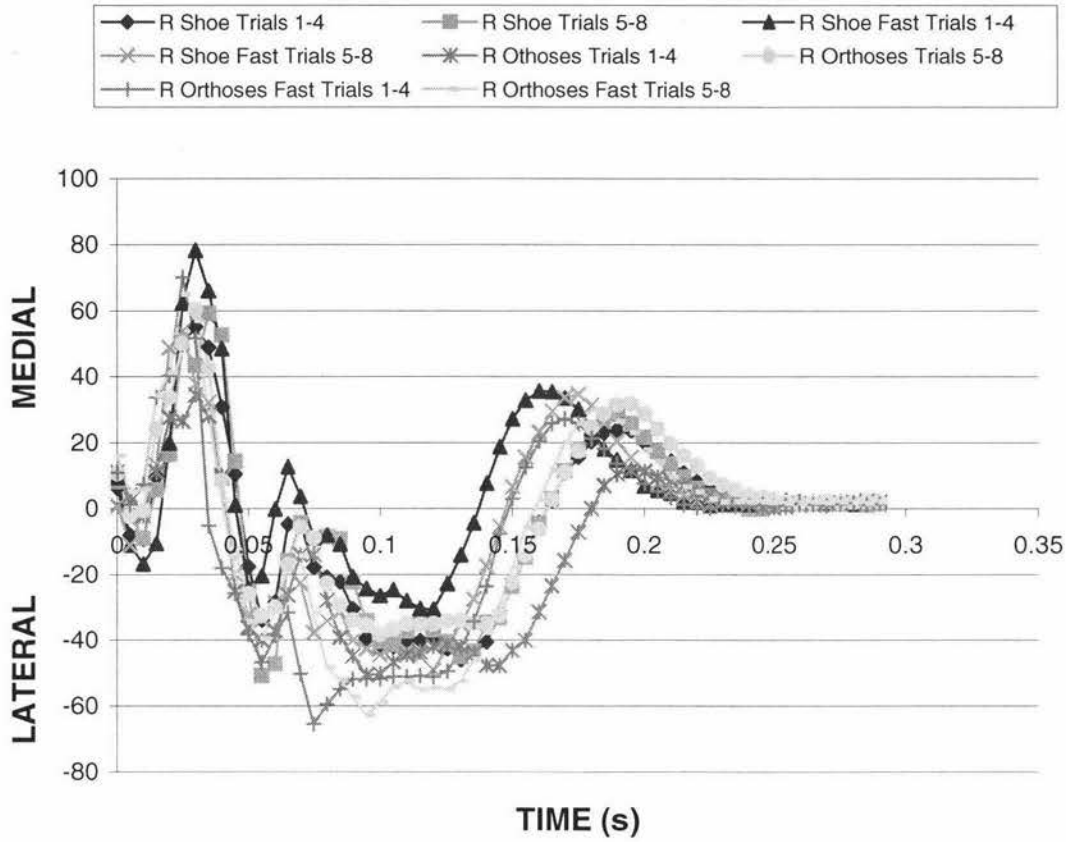

Figure 24 Mean Mediolateral Ground Reaction Force for Right Foot

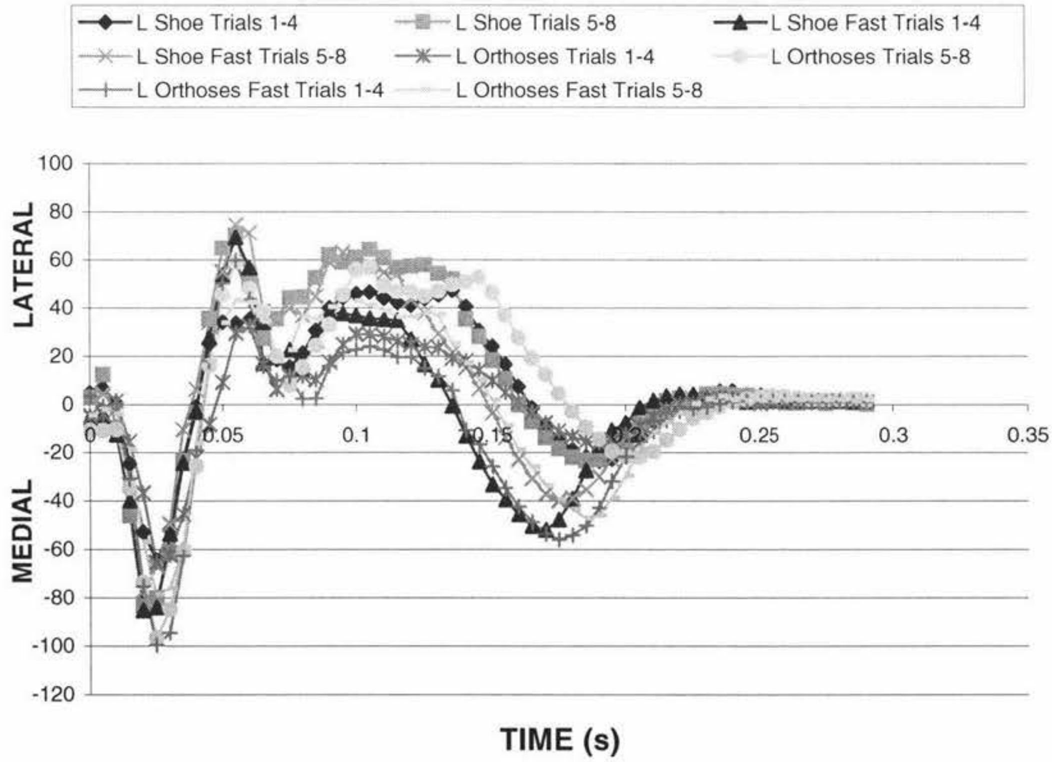

Figure 25 Mean Mediolateral Ground Reaction Force for Left Foot.

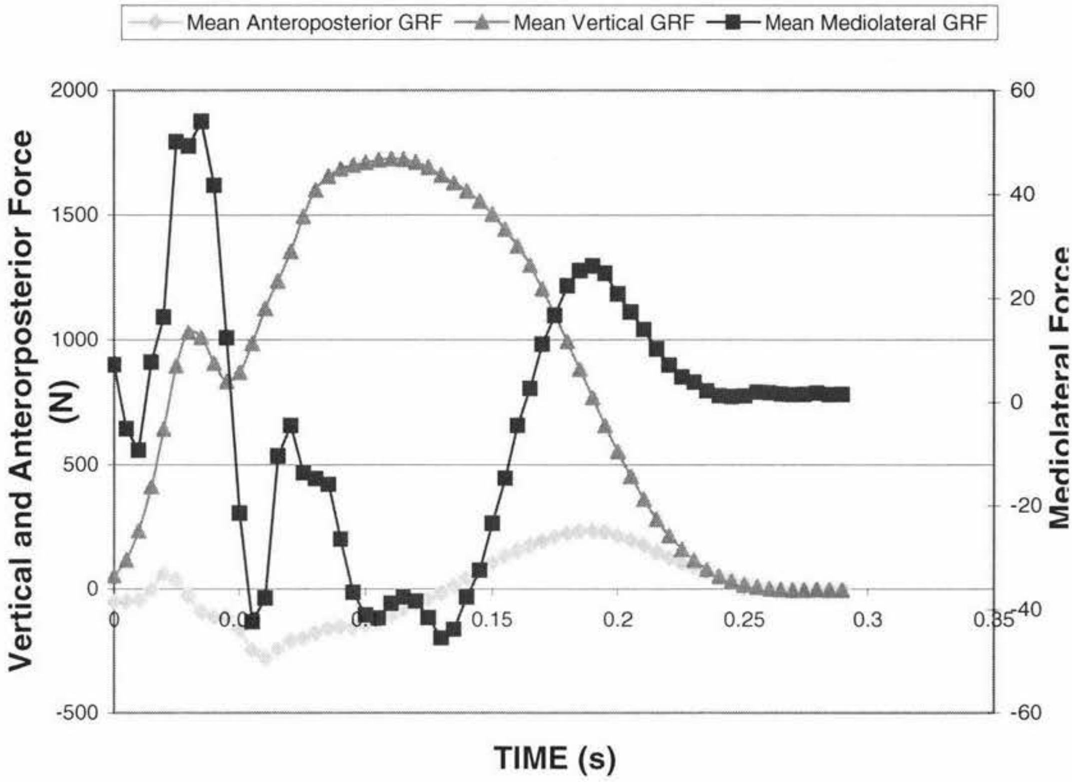

Figure 26 Mean Ground Reaction Force, Right Foot, Slow Pace, Shoe Only

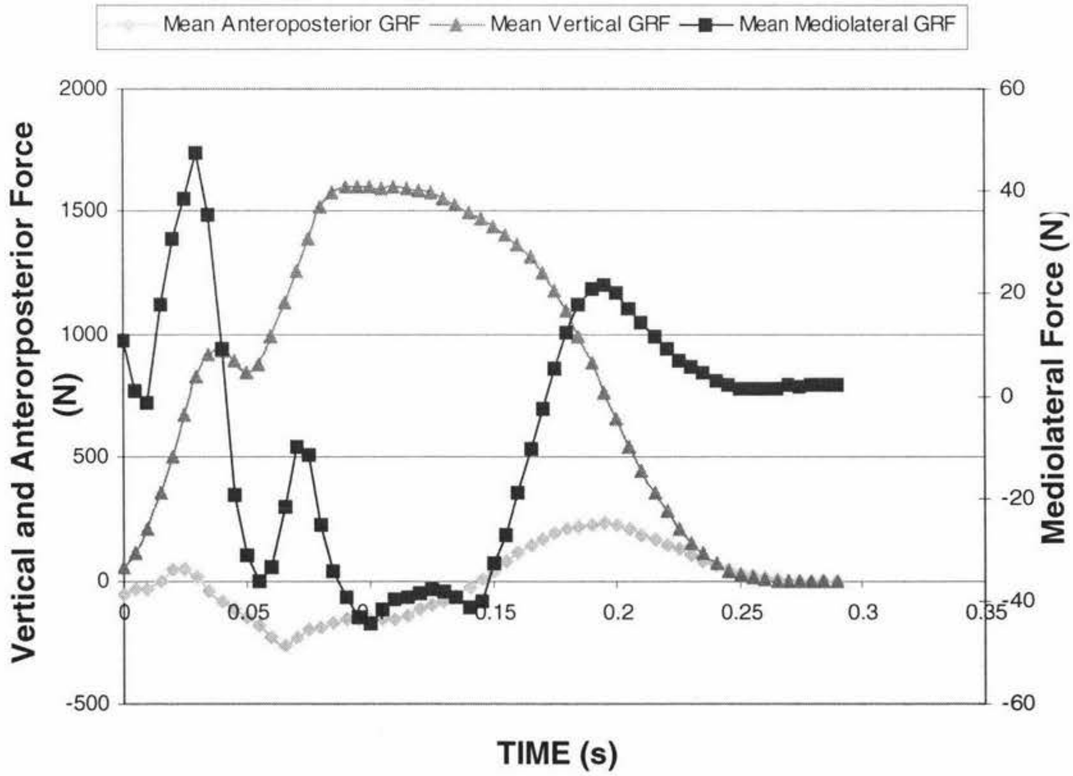

Figure 27 Mean Ground Reaction Force, Right Foot, Slow Pace, Orthoses and Shoe

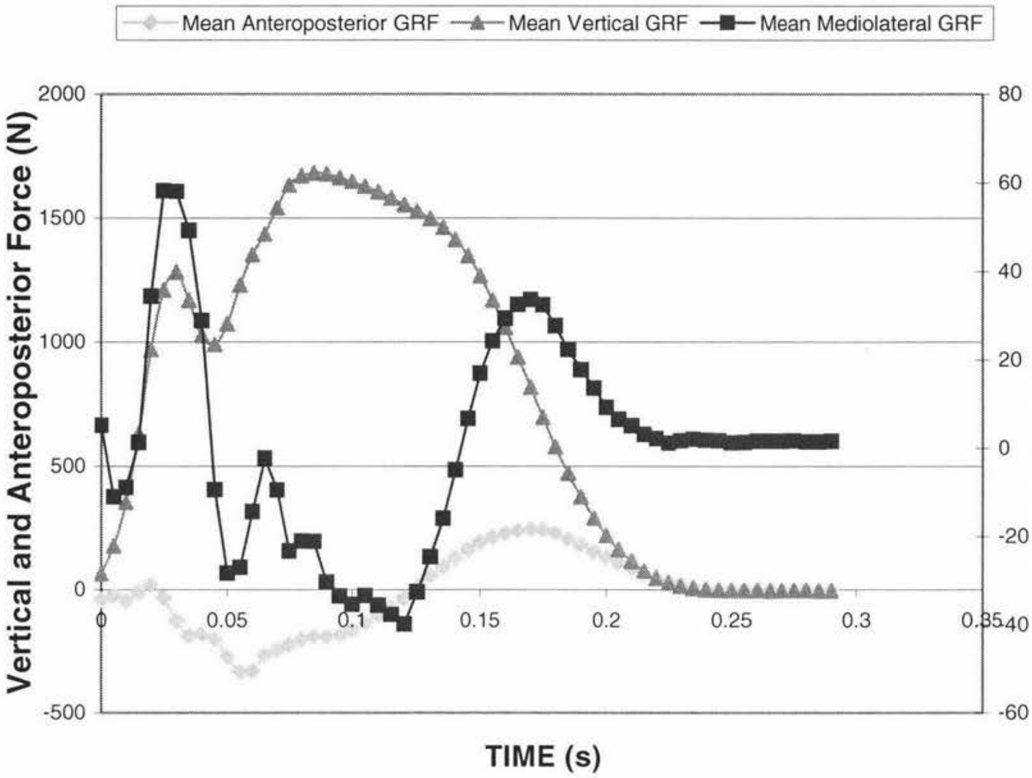

Figure 28 Mean Ground Reaction Force, Right Foot, Fast Pace, Shoe Only

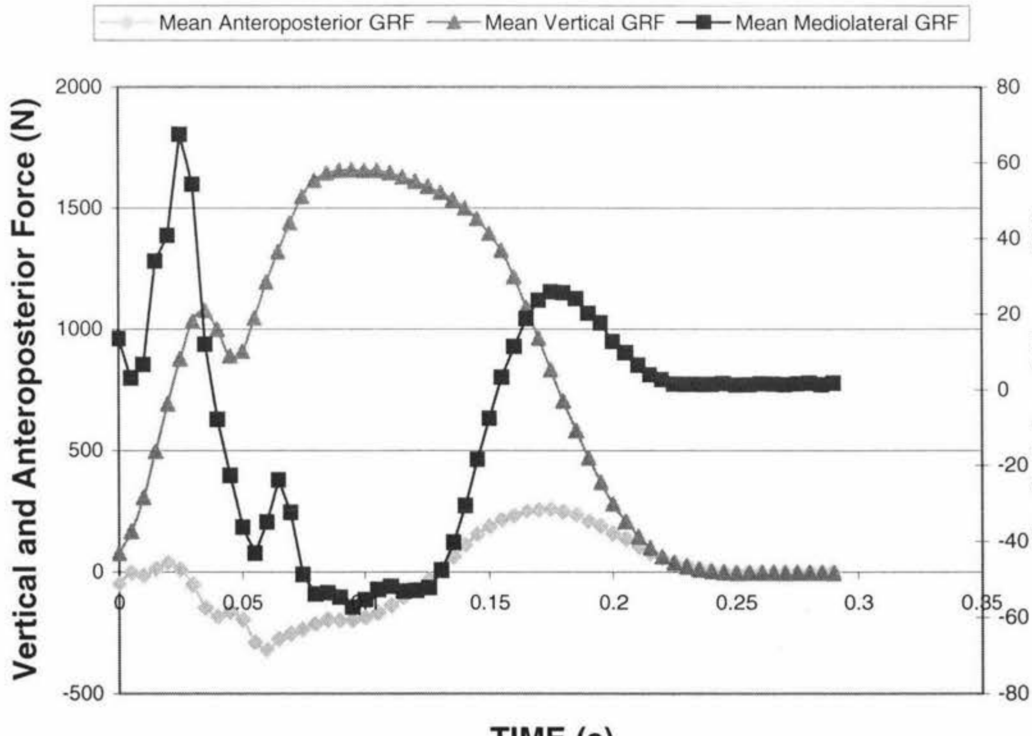

Figure 29 Mean Ground Reaction Force, Right Foot, Fast Pace, Orthoses and Shoe

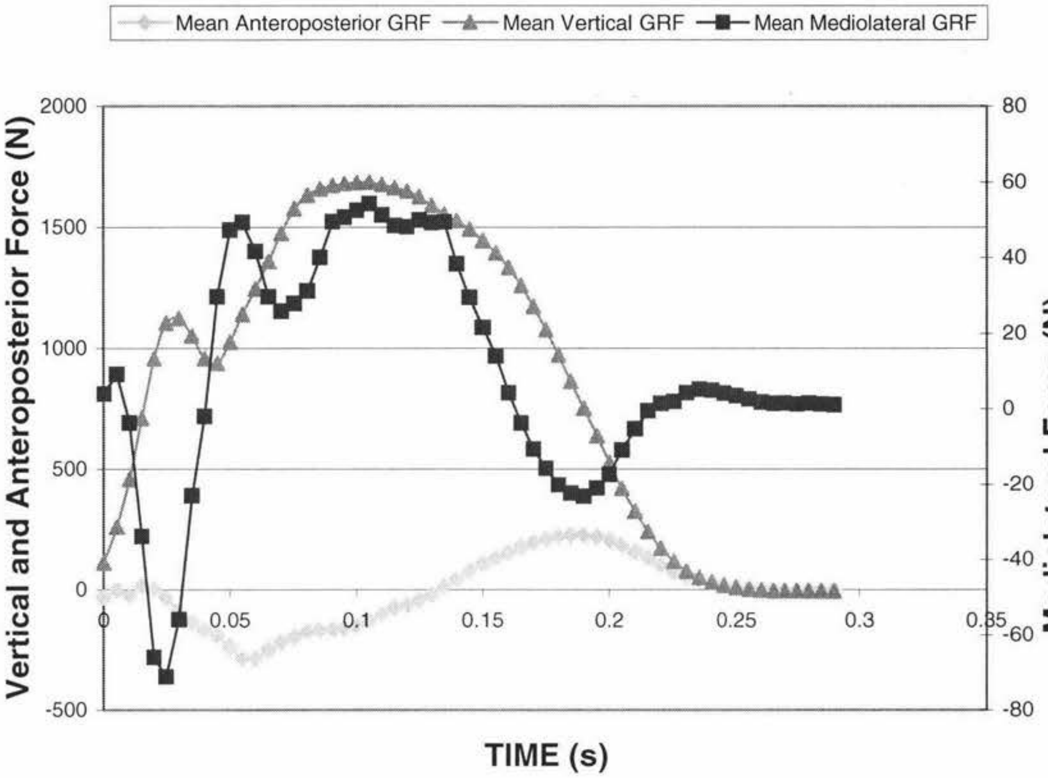

Figure 30 Mean Ground Reaction Force, Left Foot, Slow Pace, Shoe Only

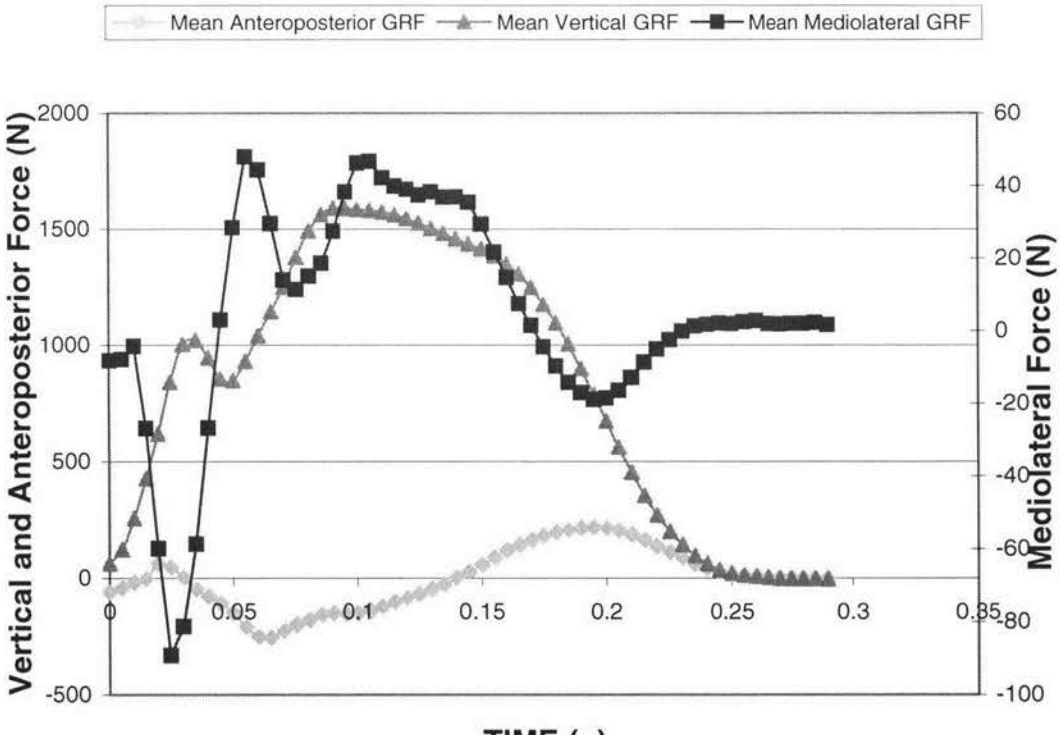

Figure 31 Mean Ground Reaction Force, Left Foot, Slow Pace, Orthoses and Shoe

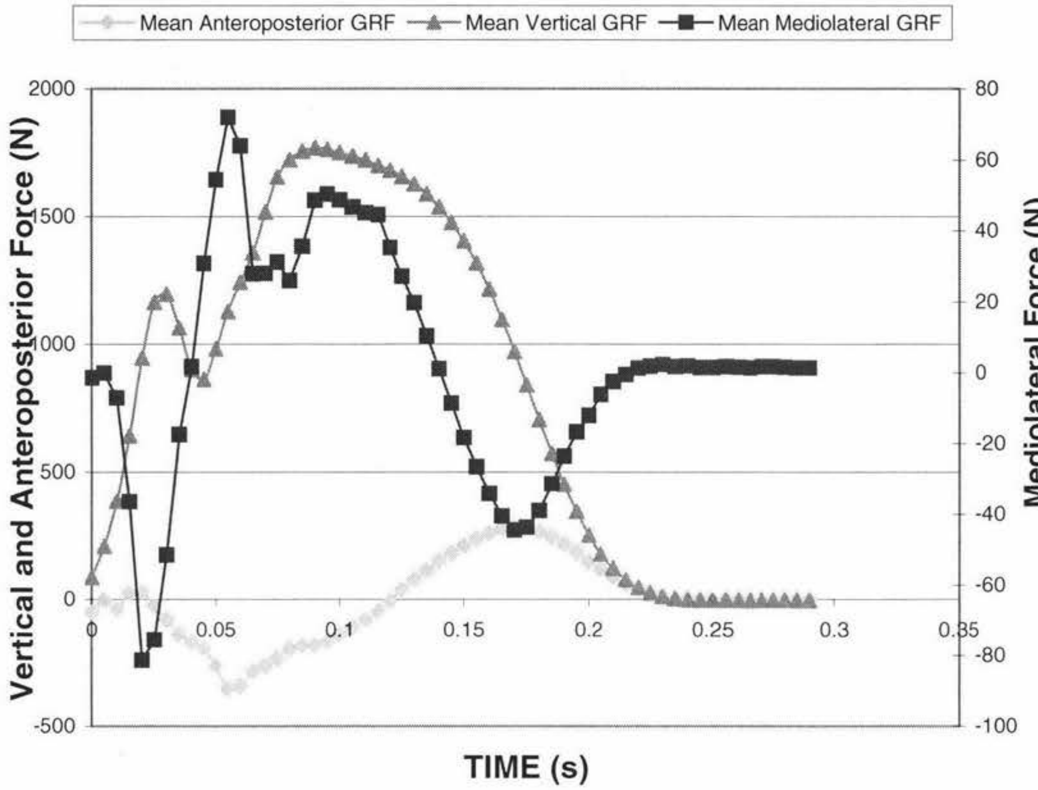

Figure 32 Mean Ground Reaction Force, Left Foot, Fast Pace, Shoe Only

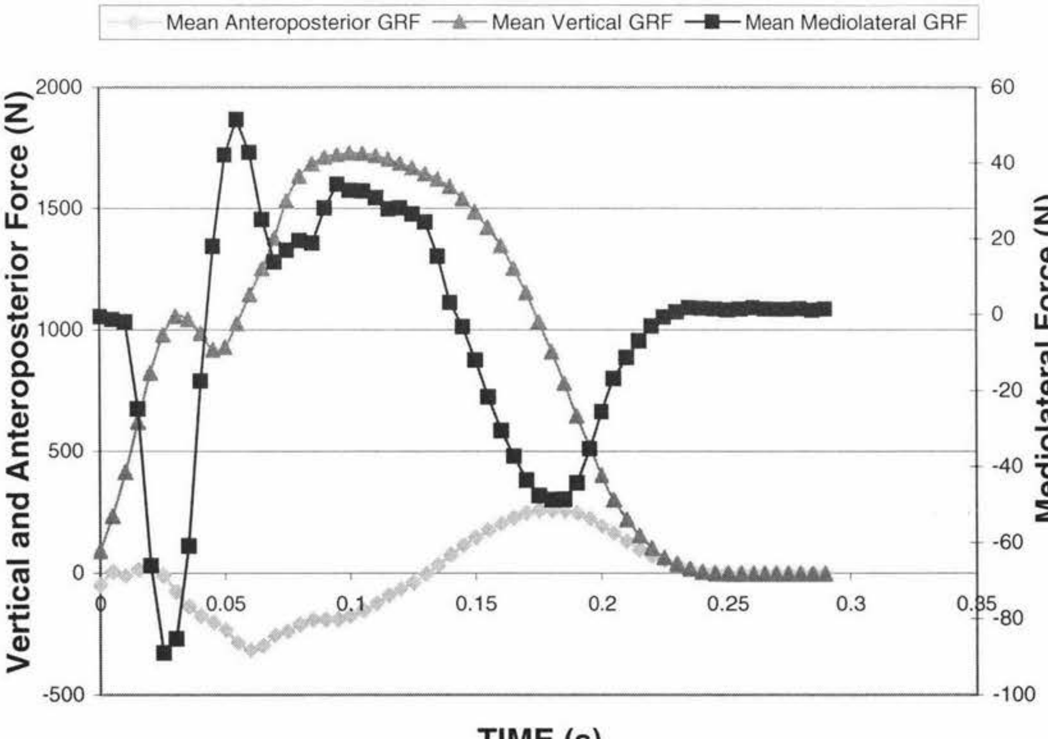

Figure 33 Mean Ground Reaction Force, Left Foot, Fast Pace, Orthoses and Shoe

Table 14 Descriptive Statistics

Key

VPK1	**P1 for Vertical Ground Reaction Force.**
VPK2	P2 for Vertical Ground Reaction Force.
Vat APO	PAP0 for Vertical Ground Reaction Force.
VTPK1	T1 for Vertical Ground Reaction Force.
VTPK2	T2 for Vertical Ground Reaction Force.
AP PK1	P1 for Anteroposterior Ground Reaction Force.
AP T PK1	T1 for Anteroposterior Ground Reaction Force.
AP T Zero Cross	T0 for Anteroposterior Ground Reaction Force.
MLP1 abs	P1 for Mediolateral Ground Reaction Force (Absolute Values).
MLP2 abs	P2 for Mediolateral Ground Reaction Force (Absolute Values).
MAP0 abs	PAP0 for Mediolateral Ground Reaction Force (Absolute Values).
MLT PK1	T1 for Mediolateral Ground Reaction Force (Absolute Values).
MLT PK2	T2 for Mediolateral Ground Reaction Force (Absolute Values).

General Linear Model (Force Parameters)

Notes

Between-Subjects Factors

		Value Label	N
Condition	1	No Orthoses	32
	2	Orthoses	32
Slow/Fast	1	Slow	32
	2	Fast	32

Descriptive Statistics

	Condition	Slow/Fast	Mean	Std. Deviation	N
AP Pk 1	No Orthoses	Slow	-296.1805	22.71591	16
		Fast	-361.4100	20.68111	16
		Total	-328.7953	39.42935	32
	Orthoses	Slow	-266.2725	24.11636	16
		Fast	-329.1019	18.29897	16
		Total	-297.6872	38.23822	32
	Total	Slow	-281.2265	27.60323	32
		Fast	-345.2559	25.26564	32
		Total	-313.2412	41.59619	64
V Pk 1	No Orthoses	Slow	1107.1694	132.40797	16
		Fast	1253.4706	98.11530	16
		Total	1180.3200	136.61929	32
	Orthoses	Slow	985.0900	121.42095	16
		Fast	1079.5106	136.81240	16
		Total	1032.3003	135.98295	32
	Total	Slow	1046.1297	139.50960	32
		Fast	1166.4906	146.71239	32
		Total	1106.3102	154.42670	64
V Pk 2	No Orthoses	Slow	1727.0267	107.11966	16
		Fast	1745.9344	143.29060	16
		Total	1736.4805	124.81758	32
	Orthoses	Slow	1623.5306	91.82527	16
		Fast	1728.5600	118.89212	16
		Total	1676.0453	117.33019	32
	Total	Slow	1675.2787	111.33921	32
		Fast	1737.2472	129.81724	32
		Total	1706.2629	123.96611	64

V at AP0	No Orthoses	Slow	1600.9270	144.84170	16
		Fast	1606.3656	174.75436	16
		Total	1603.6463	157.91072	32
	Orthoses	Slow	1464.2888	125.66213	16
		Fast	1607.6038	182.06922	16
		Total	1535.9462	170.23855	32
	Total	Slow	1532.6079	150.36636	32
		Fast	1606.9847	175.54860	32
		Total	1569.7963	166.41689	64
MLP1abs	No Orthoses	Slow	70.2274	16.46961	16
		Fast	83.3931	23.49704	16
		Total	76.8103	21.05068	32
	Orthoses	Slow	70.6838	27.28655	16
		Fast	79.1250	19.43302	16
		Total	74.9044	23.69361	32
	Total	Slow	70.4556	22.17141	32
		Fast	81.2591	21.32092	32
		Total	75.8573	22.25330	64
MLP2abs	No Orthoses	Slow	60.1114	31.90846	16
		Fast	57.6844	30.00957	16
		Total	58.8979	30.49484	32
	Orthoses	Slow	47.0688	19.19782	16
		Fast	48.4462	18.44833	16
		Total	47.7575	18.53387	32
	Total	Slow	53.5901	26.73736	32
		Fast	53.0653	24.94929	32
		Total	53.3277	25.65410	64
MAPOabs	No Orthoses	Slow	45.6200	28.67590	16
		Fast	39.4325	20.56530	16
		Total	42.5263	24.74702	32
	Orthoses	Slow	42.2681	20.62095	16
		Fast	39.2313	21.54001	16
		Total	40.7497	20.79990	32
	Total	Slow	43.9441	24.62810	32
		Fast	39.3319	20.71611	32
		Total	41.6380	22.69434	64

Multivariate Tests(b)

Effect		Value	F	Hypothesis df	Error df	Sig.
Intercept	Pillai's Trace	.998	3895.324(a)	7.000	54.000	.000
	Wilks' Lambda	.002	3895.324(a)	7.000	54.000	.000
	Hotelling's Trace	504.949	3895.324(a)	7.000	54.000	.000
	Roy's Largest Root	504.949	3895.324(a)	7.000	54.000	.000
conditio	Pillai's Trace	.492	7.478(a)	7.000	54.000	.000
	Wilks' Lambda	.508	7.478(a)	7.000	54.000	.000
	Hotelling's Trace	.969	7.478(a)	7.000	54.000	.000
	Roy's Largest Root	.969	7.478(a)	7.000	54.000	.000
speed_gr	Pillai's Trace	.760	24.449(a)	7.000	54.000	.000
	Wilks' Lambda	.240	24.449(a)	7.000	54.000	.000
	Hotelling's Trace	3.169	24.449(a)	7.000	54.000	.000
	Roy's Largest Root	3.169	24.449(a)	7.000	54.000	.000
conditio * speed_gr	Pillai's Trace	.058	.476(a)	7.000	54.000	.848
	Wilks' Lambda	.942	.476(a)	7.000	54.000	.848
	Hotelling's Trace	.062	.476(a)	7.000	54.000	.848
	Roy's Largest Root	.062	.476(a)	7.000	54.000	.848

a Exact statistic

b Design: Intercept+conditio+speed_gr+conditio * speed_gr

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	AP Pk 1	81102.710(a)	3	27034.237	58.133	.000
	V Pk 1	593111.728(b)	3	197703.909	13.046	.000
	V Pk 2	149548.019(c)	3	49849.340	3.654	.017
	V at AP0	237882.883(d)	3	79294.294	3.157	.031
	MLP1abs	2014.845(e)	3	671.615	1.381	.257
	MLP2abs	2048.042(f)	3	682.681	1.039	.382
	MAPOabs	430.561(g)	3	143.520	.269	.848
Intercept	AP Pk 1	6279684.270	1	6279684.270	13503.450	.000
	V Pk 1	78331018.357	1	78331018.357	5168.730	.000
	V Pk 2	186325322.636	1	186325322.636	13656.701	.000
	V at AP0	157712661.542	1	157712661.542	6279.722	.000
	MLP1abs	368277.326	1	368277.326	757.166	.000
	MLP2abs	182005.996	1	182005.996	277.066	.000
	MAPOabs	110958.108	1	110958.108	207.939	.000
conditio	AP Pk 1	15483.403	1	15483.403	33.295	.000
	V Pk 1	350557.246	1	350557.246	23.132	.000
	V Pk 2	58438.659	1	58438.659	4.283	.043
	V at AP0	73332.737	1	73332.737	2.920	.093
	MLP1abs	58.119	1	58.119	.119	.731
	MLP2abs	1985.737	1	1985.737	3.023	.087
	MAPOabs	50.499	1	50.499	.095	.759
speed_gr	AP Pk 1	65596.265	1	65596.265	141.054	.000
	V Pk 1	231788.084	1	231788.084	15.295	.000
	V Pk 2	61441.573	1	61441.573	4.503	.038
	V at AP0	88510.606	1	88510.606	3.524	.065
	MLP1abs	1867.444	1	1867.444	3.839	.055
	MLP2abs	4.406	1	4.406	.007	.935
	MAPOabs	340.356	1	340.356	.638	.428
conditio * speed_gr	AP Pk 1	23.042	1	23.042	.050	.825
	V Pk 1	10766.397	1	10766.397	.710	.403
	V Pk 2	29667.786	1	29667.786	2.174	.146
	V at AP0	76039.540	1	76039.540	3.028	.087
	MLP1abs	89.282	1	89.282	.184	.670
	MLP2abs	57.899	1	57.899	.088	.768
	MAPOabs	39.706	1	39.706	.074	.786
Error	AP Pk 1	27902.578	60	465.043		
	V Pk 1	909287.439	60	15154.791		
	V Pk 2	818610.527	60	13643.509		
	V at AP0	1506875.677	60	25114.595		
	MLP1abs	29183.346	60	486.389		
	MLP2abs	39414.324	60	656.905		
	MAPOabs	32016.519	60	533.609		

Total	AP Pk 1	6388689.557	64			
	V Pk 1	79833417.524	64			
	V Pk 2	187293481.182	64			
	V at AP0	159457420.102	64			
	MLP1abs	399475.517	64			
	MLP2abs	223468.361	64			
	MAPOabs	143405.189	64			
Corrected Total	AP Pk 1	109005.288	63			
	V Pk 1	1502399.167	63			
	V Pk 2	968158.546	63			
	V at AP0	1744758.560	63			
	MLP1abs	31198.191	63			
	MLP2abs	41462.365	63			
	MAPOabs	32447.080	63			

a R Squared = .744 (Adjusted R Squared = .731)
b R Squared = .395 (Adjusted R Squared = .365)
c R Squared = .154 (Adjusted R Squared = .112)
d R Squared = .136 (Adjusted R Squared = .093)
e R Squared = .065 (Adjusted R Squared = .018)
f R Squared = .049 (Adjusted R Squared = .002)
g R Squared = .013 (Adjusted R Squared = -.036)

Custom Hypothesis Tests Index

1	Contrast Coefficients (L' Matrix)	Simple Contrast (reference category = 1) for Condition
	Transformation Coefficients (M Matrix)	Identity Matrix
	Contrast Results (K Matrix)	Zero Matrix
2	Contrast Coefficients (L' Matrix)	Simple Contrast (reference category = 1) for Slow/Fast
	Transformation Coefficients (M Matrix)	Identity Matrix
	Contrast Results (K Matrix)	Zero Matrix

Custom Hypothesis Tests #1

Contrast Results (K Matrix)

Condition Simple Contrast(a)			Dependent Variable						
			AP Pk 1	V Pk 1	V Pk 2	V at APO	MLP1abs	MLP2abs	MAPOabs
Level 2 vs. Level 1	Contrast Estimate		31.108	-148.020	-60.435	-67.700	-1.906	-11.140	-1.777
	Hypothesized Value		0	0	0	0	0	0	0
	Difference (Estimate - Hypothesized)		31.108	-148.020	-60.435	-67.700	-1.906	-11.140	-1.777
	Std. Error		5.391	30.776	29.201	39.619	5.514	6.408	5.775
	Sig.		.000	.000	.043	.093	.731	.087	.759
	95% Confidence Interval for Difference	Lower Bound	20.324	-209.581	-118.847	-146.950	-12.935	-23.957	-13.328
		Upper Bound	41.892	-86.458	-2.024	11.550	9.123	1.677	9.775

a Reference category = 1

Multivariate Test Results

	Value	F	Hypothesis df	Error df	Sig.
Pillai's trace	.492	7.478(a)	7.000	54.000	.000
Wilks' lambda	.508	7.478(a)	7.000	54.000	.000
Hotelling's trace	.969	7.478(a)	7.000	54.000	.000
Roy's largest root	.969	7.478(a)	7.000	54.000	.000

a Exact statistic

Univariate Test Results

Source	Dependent Variable	Sum of Squares	df	Mean Square	F	Sig.
Contrast	AP Pk 1	15483.403	1	15483.403	33.295	.000
	V Pk 1	350557.246	1	350557.246	23.132	.000
	V Pk 2	58438.659	1	58438.659	4.283	.043
	V at AP0	73332.737	1	73332.737	2.920	.093
	MLP1abs	58.119	1	58.119	.119	.731
	MLP2abs	1985.737	1	1985.737	3.023	.087
	MAPOabs	50.499	1	50.499	.095	.759
Error	AP Pk 1	27902.578	60	465.043		
	V Pk 1	909287.439	60	15154.791		
	V Pk 2	818610.527	60	13643.509		
	V at AP0	1506875.677	60	25114.595		
	MLP1abs	29183.346	60	486.389		
	MLP2abs	39414.324	60	656.905		
	MAPOabs	32016.519	60	533.609		

Custom Hypothesis Tests #2

Contrast Results (K Matrix)

Slow/Fast Simple Contrast(a)			Dependent Variable						
			AP Pk 1	V Pk 1	V Pk 2	V at APO	MLP1abs	MLP2abs	MAPOabs
Level 2 vs. Level 1	Contrast Estimate		-64.029	120.361	61.969	74.377	10.803	-.525	-4.612
	Hypothesized Value		0	0	0	0	0	0	0
	Difference (Estimate - Hypothesized)		-64.029	120.361	61.969	74.377	10.803	-.525	-4.612
	Std. Error		5.391	30.776	29.201	39.619	5.514	6.408	5.775
	Sig.		.000	.000	.038	.065	.055	.935	.428
	95% Confidence Interval for Difference	Lower Bound	-74.813	58.799	3.557	-4.873	-.225	-13.342	-16.164
		Upper Bound	-53.245	181.923	120.380	153.627	21.832	12.292	6.940

a Reference category = 1

Multivariate Test Results

	Value	F	Hypothesis df	Error df	Sig.
Pillai's trace	.760	24.449(a)	7.000	54.000	.000
Wilks' lambda	.240	24.449(a)	7.000	54.000	.000
Hotelling's trace	3.169	24.449(a)	7.000	54.000	.000
Roy's largest root	3.169	24.449(a)	7.000	54.000	.000

a Exact statistic

Univariate Test Results

Source	Dependent Variable	Sum of Squares	df	Mean Square	F	Sig.
Contrast	AP Pk 1	65596.265	1	65596.265	141.054	.000
	V Pk 1	231788.084	1	231788.084	15.295	.000
	V Pk 2	61441.573	1	61441.573	4.503	.038
	V at AP0	88510.606	1	88510.606	3.524	.065
	MLP1abs	1867.444	1	1867.444	3.839	.055
	MLP2abs	4.406	1	4.406	.007	.935
	MAPOabs	340.356	1	340.356	.638	.428
Error	AP Pk 1	27902.578	60	465.043		
	V Pk 1	909287.439	60	15154.791		
	V Pk 2	818610.527	60	13643.509		
	V at AP0	1506875.677	60	25114.595		
	MLP1abs	29183.346	60	486.389		
	MLP2abs	39414.324	60	656.905		
	MAPOabs	32016.519	60	533.609		

General Linear Model (Time Parameters)

Notes

Output Created		04-OCT-2004 12:32:01
Comments		
Input	Data	C:\My Documents\Teaching\Postgraduate\John Sloane MPhil\Raw Parameters 2.sav
	Filter	<none>
	Weight	<none>
	Split File	<none>
	N of Rows in Working Data File	64
Missing Value Handling	Definition of Missing	User-defined missing values are treated as missing.
	Cases Used	Statistics are based on all cases with valid data for all variables in the model.
Syntax		GLM mlt1 mlt2 apt1 apt0 vt1 vt2 BY conditio speed_gr /CONTRAST (conditio)=Simple(1) /CONTRAST (speed_gr)=Simple(1) /METHOD = SSTYPE(3) /INTERCEPT = INCLUDE /PRINT = DESCRIPTIVE /CRITERIA = ALPHA(.05) /DESIGN = conditio speed_gr conditio*speed_gr .
Resources	Elapsed Time	0:00:00.08

Between-Subjects Factors

		Value Label	N
Condition	1	No Orthoses	32
	2	Orthoses	32
Slow/Fast	1	Slow	32
	2	Fast	32

Descriptive Statistics

	Condition	Slow/Fast	Mean	Std. Deviation	N
ML T Pk1	No Orthoses	Slow	.02741	.005826	16
		Fast	.02438	.004425	16
		Total	.02589	.005317	32
	Orthoses	Slow	.02750	.004830	16
		Fast	.02563	.002500	16
		Total	.02656	.003902	32
	Total	Slow	.02746	.005264	32
		Fast	.02500	.003592	32
		Total	.02623	.004639	64
ML T Pk 2	No Orthoses	Slow	.05549	.006059	16
		Fast	.05562	.004031	16
		Total	.05556	.005063	32
	Orthoses	Slow	.05594	.004171	16
		Fast	.05250	.008756	16
		Total	.05422	.006969	32
	Total	Slow	.05571	.005122	32
		Fast	.05406	.006891	32
		Total	.05489	.006080	64
AP T Pk 1	No Orthoses	Slow	.05866	.003399	16
		Fast	.05719	.003637	16
		Total	.05792	.003543	32
	Orthoses	Slow	.06469	.002213	16
		Fast	.06063	.003096	16
		Total	.06266	.003356	32
	Total	Slow	.06167	.004163	32
		Fast	.05891	.003753	32
		Total	.06029	.004172	64
AP T zero cross	No Orthoses	Slow	.1168	.06725	16
		Fast	.1225	.00577	16
		Total	.1196	.04704	32
	Orthoses	Slow	.1422	.00752	16
		Fast	.1291	.00491	16
		Total	.1356	.00914	32
	Total	Slow	.1295	.04881	32
		Fast	.1258	.00624	32
		Total	.1276	.03457	64
V T Pk 1	No Orthoses	Slow	.02951	.004315	16
		Fast	.02875	.002887	16
		Total	.02913	.003632	32
	Orthoses	Slow	.03656	.005072	16
		Fast	.03344	.004366	16
		Total	.03500	.004919	32
	Total	Slow	.03304	.005857	32
		Fast	.03109	.004350	32
		Total	.03206	.005210	64

V T Pk 2	No Orthoses	Slow	.1035	.01397	16
		Fast	.0884	.00908	16
		Total	.0960	.01390	32
	Orthoses	Slow	.0959	.01294	16
		Fast	.0975	.01342	16
		Total	.0967	.01299	32
	Total	Slow	.0997	.01380	32
		Fast	.0930	.01217	32
		Total	.0964	.01335	64

Multivariate Tests(b)

Effect		Value	F	Hypothesis df	Error df	Sig.
Intercept	Pillai's Trace	.998	4994.255(a)	6.000	55.000	.000
	Wilks' Lambda	.002	4994.255(a)	6.000	55.000	.000
	Hotelling's Trace	544.828	4994.255(a)	6.000	55.000	.000
	Roy's Largest Root	544.828	4994.255(a)	6.000	55.000	.000
Conditio	Pillai's Trace	.553	11.335(a)	6.000	55.000	.000
	Wilks' Lambda	.447	11.335(a)	6.000	55.000	.000
	Hotelling's Trace	1.237	11.335(a)	6.000	55.000	.000
	Roy's Largest Root	1.237	11.335(a)	6.000	55.000	.000
speed_gr	Pillai's Trace	.289	3.717(a)	6.000	55.000	.004
	Wilks' Lambda	.711	3.717(a)	6.000	55.000	.004
	Hotelling's Trace	.406	3.717(a)	6.000	55.000	.004
	Roy's Largest Root	.406	3.717(a)	6.000	55.000	.004
conditio * speed_gr	Pillai's Trace	.194	2.209(a)	6.000	55.000	.056
	Wilks' Lambda	.806	2.209(a)	6.000	55.000	.056
	Hotelling's Trace	.241	2.209(a)	6.000	55.000	.056
	Roy's Largest Root	.241	2.209(a)	6.000	55.000	.056

a Exact statistic
b Design: Intercept+conditio+speed_gr+conditio * speed_gr

Tests of Between-Subjects Effects

Source	Dependent Variable	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	ML T Pk1	.000(a)	3	3.634E-05	1.749	.167
	ML T Pk 2	.000(b)	3	4.112E-05	1.119	.349
	AP T Pk 1	.001(c)	3	.000	17.240	.000
	AP T zero cross	.006(d)	3	.002	1.647	.188
	V T Pk 1	.001(e)	3	.000	11.785	.000
	V T Pk 2	.002(f)	3	.001	3.946	.012

Intercept	ML T Pk1	.044	1	.044	2119.040	.000
	ML T Pk 2	.193	1	.193	5245.668	.000
	AP T Pk 1	.233	1	.233	23699.191	.000
	AP T zero cross	1.043	1	1.043	899.386	.000
	V T Pk 1	.066	1	.066	3668.610	.000
	V T Pk 2	.594	1	.594	3802.523	.000
Conditio	ML T Pk1	7.175E-06	1	7.175E-06	.345	.559
	ML T Pk 2	2.870E-05	1	2.870E-05	.781	.380
	AP T Pk 1	.000	1	.000	36.500	.000
	AP T zero cross	.004	1	.004	3.526	.065
	V T Pk 1	.001	1	.001	30.743	.000
	V T Pk 2	8.681E-06	1	8.681E-06	.056	.814
speed_gr	ML T Pk1	9.646E-05	1	9.646E-05	4.643	.035
	ML T Pk 2	4.365E-05	1	4.365E-05	1.188	.280
	AP T Pk 1	.000	1	.000	12.487	.001
	AP T zero cross	.000	1	.000	.190	.665
	V T Pk 1	6.034E-05	1	6.034E-05	3.364	.072
	V T Pk 2	.001	1	.001	4.684	.034
conditio * speed_gr	ML T Pk1	5.389E-06	1	5.389E-06	.259	.612
	ML T Pk 2	5.102E-05	1	5.102E-05	1.388	.243
	AP T Pk 1	2.682E-05	1	2.682E-05	2.732	.104
	AP T zero cross	.001	1	.001	1.225	.273
	V T Pk 1	2.239E-05	1	2.239E-05	1.248	.268
	V T Pk 2	.001	1	.001	7.098	.010
Error	ML T Pk1	.001	60	2.078E-05		
	ML T Pk 2	.002	60	3.676E-05		
	AP T Pk 1	.001	60	9.816E-06		
	AP T zero cross	.070	60	.001		
	V T Pk 1	.001	60	1.794E-05		
	V T Pk 2	.009	60	.000		
Total	ML T Pk1	.045	64			
	ML T Pk 2	.195	64			
	AP T Pk 1	.234	64			
	AP T zero cross	1.118	64			
	V T Pk 1	.068	64			
	V T Pk 2	.605	64			
Corrected Total	ML T Pk1	.001	63			
	ML T Pk 2	.002	63			
	AP T Pk 1	.001	63			
	AP T zero cross	.075	63			
	V T Pk 1	.002	63			
	V T Pk 2	.011	63			

a R Squared = .080 (Adjusted R Squared = .034)
b R Squared = .053 (Adjusted R Squared = .006)
c R Squared = .463 (Adjusted R Squared = .436)
d R Squared = .076 (Adjusted R Squared = .030)
e R Squared = .371 (Adjusted R Squared = .339)
f R Squared = .165 (Adjusted R Squared = .123)

Custom Hypothesis Tests Index

1	Contrast Coefficients (L' Matrix)	Simple Contrast (reference category = 1) for Condition
	Transformation Coefficients (M Matrix)	Identity Matrix
	Contrast Results (K Matrix)	Zero Matrix
2	Contrast Coefficients (L' Matrix)	Simple Contrast (reference category = 1) for Slow/Fast
	Transformation Coefficients (M Matrix)	Identity Matrix
	Contrast Results (K Matrix)	Zero Matrix

Custom Hypothesis Tests #1

Contrast Results (K Matrix)

Condition Simple Contrast(a)			Dependent Variable					
			ML T Pk1	ML T Pk 2	AP T Pk 1	AP T zero cross	V T Pk 1	V T Pk 2
Level 2 vs. Level 1	Contrast Estimate		.001	-.001	.005	.016	.006	.001
	Hypothesized Value		0	0	0	0	0	0
	Difference (Estimate - Hypothesized)		.001	-.001	.005	.016	.006	.001
	Std. Error		.001	.002	.001	.009	.001	.003
	Sig.		.559	.380	.000	.065	.000	.814
	95% Confidence Interval for Difference	Lower Bound	-.002	-.004	.003	-.001	.004	-.006
		Upper Bound	.003	.002	.006	.033	.008	.007

a Reference category = 1

Multivariate Test Results

	Value	F	Hypothesis df	Error df	Sig.
Pillai's trace	.553	11.335(a)	6.000	55.000	.000
Wilks' lambda	.447	11.335(a)	6.000	55.000	.000
Hotelling's trace	1.237	11.335(a)	6.000	55.000	.000
Roy's largest root	1.237	11.335(a)	6.000	55.000	.000

a Exact statistic

Univariate Test Results

Source	Dependent Variable	Sum of Squares	df	Mean Square	F	Sig.
Contrast	ML T Pk1	.000	1	.000	.345	.559
	ML T Pk 2	.000	1	.000	.781	.380
	AP T Pk 1	.000	1	.000	36.500	.000
	AP T zero cross	.004	1	.004	3.526	.065
	V T Pk 1	.001	1	.001	30.743	.000
	V T Pk 2	.000	1	.000	.056	.814
Error	ML T Pk1	.001	60	.000		
	ML T Pk 2	.002	60	.000		
	AP T Pk 1	.001	60	.000		
	AP T zero cross	.070	60	.001		
	V T Pk 1	.001	60	.000		
	V T Pk 2	.009	60	.000		

Custom Hypothesis Tests #2

Contrast Results (K Matrix)

Slow/Fast Simple Contrast(a)			Dependent Variable					
			ML T Pk1	ML T Pk 2	AP T Pk 1	AP T zero cross	V T Pk 1	V T Pk 2
Level 2 vs. Level 1	Contrast Estimate		-.002	-.002	-.003	-.004	-.002	-.007
	Hypothesized Value		0	0	0	0	0	0
	Difference (Estimate - Hypothesized)		-.002	-.002	-.003	-.004	-.002	-.007
	Std. Error		.001	.002	.001	.009	.001	.003
	Sig.		.035	.280	.001	.665	.072	.034
	95% Confidence Interval for Difference	Lower Bound	-.005	-.005	-.004	-.021	-.004	-.013
		Upper Bound	.000	.001	-.001	.013	.000	-.001

a Reference category = 1

Multivariate Test Results

	Value	F	Hypothesis df	Error df	Sig.
Pillai's trace	.289	3.717(a)	6.000	55.000	.004
Wilks' lambda	.711	3.717(a)	6.000	55.000	.004
Hotelling's trace	.406	3.717(a)	6.000	55.000	.004
Roy's largest root	.406	3.717(a)	6.000	55.000	.004

a Exact statistic

Univariate Test Results

Source	Dependent Variable	Sum of Squares	df	Mean Square	F	Sig.
Contrast	ML T Pk1	.000	1	.000	4.643	.035
	ML T Pk 2	.000	1	.000	1.188	.280
	AP T Pk 1	.000	1	.000	12.487	.001
	AP T zero cross	.000	1	.000	.190	.665
	V T Pk 1	.000	1	.000	3.364	.072
	V T Pk 2	.001	1	.001	4.684	.034
Error	ML T Pk1	.001	60	.000		
	ML T Pk 2	.002	60	.000		
	AP T Pk 1	.001	60	.000		
	AP T zero cross	.070	60	.001		
	V T Pk 1	.001	60	.000		
	V T Pk 2	.009	60	.000		