Effects of Footwear Choice amongst Field Hockey Participants on Factors Influencing Overuse Injuries

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Abstract

Field hockey is a popular sport played worldwide. Due to the demands of the game, injuries are common, restricting participation. Injuries occur due to a single traumatic event or are due to repetitive loading. Injuries caused by repetitive loading, known as overuse injuries, have been linked to various measureable kinetic and kinematic variables. The magnitude, direction and distribution of the applied loads have all been identified as factors influencing the onset of an injury. Furthermore, footwear, surface and speed of locomotion have also been identified as factors which will influence injuries. Altering these variables could assist in reducing the prevalence of overuse injuries across a population of participants.

The initial study within this research investigated the magnitude and direction of applied ground reaction forces to the alignment of the tibia. Testing different insole surfaces, a rougher surface was found to increase proprioception, identified through a significant difference in the alignment of the tibia to the resultant ground reaction force vector. The next investigation used an adapted plantar pressure measuring device to record peak pressures between the uppers of the foot and shoes during various sports specific movements. Levels of peak pressures were found to match those under the feet. This method of assessment is therefore recommended for testing footwear designs in the future.

Whilst there is an established relationship between the Ground Reaction Force (GRF) and tibial acceleration, this study compared a variety of previously reported GRF characteristics such as loading rates, peak forces and time to peak forces, to tibial accelerations using a shank mounted accelerometer system developed for this research. This provided identification of key ground reaction force characteristics linked to impact shock, for assessment of the footwear.

This study then assessed a set of different footwear typically used by field hockey participants. The influence of these shoes on GRFs, and in–shoe pressure was investigated during running and jogging. These shoes along with a new prototype of running shoe designed to encourage forefoot running, were also assessed for their influence on impact shock measured directly using the shank mounted accelerometer system. A custom made computer program was employed to analyse the data. This program could be used in future research and clinical assessment.

The results of the footwear evaluation identified that moulded cleat designs with a lack of midsole cushioning exposed participants to injury causing loading of the musculoskeletal system and therefore were not recommended for use in field hockey participation. Furthermore, the prototype running shoes were adjudged to require pre-training and further assessment. The other shoes which included running, soccer and hockey specific footwear did not produce any significant differences across the population of participants. However it was found that individual assessment produced many differences between the shoes. These results demonstrated that the shoes can have a positive and negative effect for different individuals on kinetics linked to overuse injuries. It was concluded that individual assessment was needed for identification of the correct footwear choice.

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Abbreviations

Acceleration Due to Earth's Gravity	g
Analysis of Variance	ANOVA
Anterior Cruciate Ligament Injury	ACL
Average Vertical Loading Rate from Foot Down to 1st vertical Force Peak	AVLR
Average Vertical Loading Rate from 20 to 80% of Foot Down to 1st Vertical Force Peak	AVL20T80
Average Vertical Loading Rate from 20 to 90% of Foot Down to 1st Vertical Force Peak	AVL20T90
Average Vertical Loading Rate from 50N to 50N plus Bodyweight	AVL50NT50NBW
Body Weight	BW
Centimetres	cm
Centre of Pressure	CoP
Ground Reaction Force	GRF
Ground Reaction Force 'x' Component	Fx
Ground Reaction Force 'y' Component	Fy
Ground Reaction Force 'z' Component	Fz
Indoor Synthetic Sports Surface	ISSS
Kilometre	Km
Metacarpophalangeal Joint	MPJ
Metatarsal Head	MTH
Moment About the 'x' Axis	Mx
Moment About the 'y' Axis	My
Moment About the 'z' Axis	Mz
Negative	-ve
Newtons	N
Outdoor Synthetic Sports Surface	OSSS
Three Dimensional	3D

Time to 1st Vertical Ground Reaction Force Peak	TVFP1
Time to 2nd Vertical Ground Reaction Force Peak	TVFP2
Time to Peak Vertical Loading Rate	TPVLR
Kilopascals	KPa
Peak Vertical Loading Rate	PVLR
Positive	+ve
Seconds	s

Publications and Conference Presentations

Peer Reviewed Publications

Greenhalgh, A., Chockalingam, N. O'Connor, A., Dangerfield, P., & Cochrane, T. (2006) Alignment of the tibia to applied forces during a Cutting Movement. Clin Anat 19(8) 771.

Conference Oral Presentations

Greenhalgh, A. Chockalingam, N, O'Connor, A. Dangerfield, P. & Cochrane, T. (2006) Comparing Ground Reaction Force Vector to the Alignment of Tibia during a Cutting Movement. Presented at the BACA scientific meeting, Keele.

Greenhalgh, A., Chockalingam, N. & Morgan A. (2008) A comparison of in-shoe localised pressure experienced during locomotive movement over an artificial surface in various types of sports specific footwear. Presented at the Sixth Staffordshire Conference on Clinical Biomechanics, Stoke on Trent.

Greenhalgh, A. Chockalingam, N, Chatterly, F. & O'Connor, A. (2006) Pressure distribution in the uppers of shoes during sports specific movements Presented at the BASES annual conference, Wolverhampton.

Chapter One

Introduction

1 Introduction

1.1 Background to the thesis

Field hockey is a popular sport played worldwide with 127 countries affiliated with the International Hockey Federation (FIH). As with many invasion games i.e. soccer and handball, hockey is played between two teams whose main aim is to move the ball into the opponent's goal.

Hockey footwear is designed specifically for the demands of the game. Designs incorporate reinforced sides to protect the foot from ball and stick impacts, cushioning systems in the midsole to protect from foot to ground impacts, and rubber outsoles designed for traction on the synthetic sports surfaces. Nevertheless, hockey players are commonly observed wearing shoes designed for other sports i.e. soccer and running shoes which may alter the likelihood of injury and affect the performance of the participant.

In general, footwear used in field hockey participation has two main purposes, to provide protection from injury and to enhance performance. Wearing footwear as opposed to being barefoot while participating in field hockey activities is therefore the accepted practice. Research has reported detrimental loading of the musculoskeletal system in barefoot running compared to shod, in participants who habitually wear shoes (De Wit et al., 2000). However, there is evidence that through adaptation of running style, forefoot landing in barefoot conditions can produce lower loading rates in many runners than in normal rear foot shod running conditions

(Lieberman *et al.*, 2010;Oakley and Pratt, 1988). Barefoot running may be used as part of a training programme, for activities such as warm up runs or fitness work. However, the risks of injuries to participants who are not habitual barefoot runners make such a strategy hazardous. Furthermore, during field hockey participation where sticks and balls are used, the feet require protection from external impacts inherent in the sport as well as sharp objects that may be present in the outdoor surface. Using barefoot locomotion as part of training would require participant education in order to make sure the participants are running correctly and not exposing themselves to an increased risk of injury. When performing multidirectional skills barefoot conditions would detriment the performance due to insufficient friction between the feet and the synthetic sports surface. This factor alone justifies the need for footwear to be worn during field hockey matches.

The friction between footwear and the surface is an important factor to consider when participants desire the ability to change direction at high speed. Performing movements involving sudden changes in direction requires shear forces of magnitudes larger than the bodyweight (BW) of the athlete (McClay et al., 1994). Acting parallel to the ground, this frictional force known as translation force has been shown to be dependent of the footwear grip characteristics (Li and Chen, 2004). A further consideration is the rotational friction, which refers to the frictional force acting against the turning moment of the outer sole with the ground (Heidt *et al.*, 1996;Frederick, 1993). Footwear can be specifically designed to provide high translation friction, which generally enhances performance and low rotational friction, which generally protects from injury. This facilitates quick movements as well as reducing the risk of injury due to torsional forces about the long axis of the lower extremities (Wannop *et al.*, 2010;Frederick, 1993). It is clear that footwear is an important piece of equipment for both preventing injury and enhancing performance within field hockey. However, there is a paucity of information regarding the most effective footwear designs of field hockey shoes for reducing injury occurrence. By identifying the most effective footwear designs to enhance protection from injury field hockey players may be able to reduce the amount of time spent unable to participate in field hockey training and matches due to injury.

The effectiveness of the design and construction of the footwear is influenced by the surface characteristics. Field hockey was originally played on a natural grass surface. Synthetic sports surfaces are now generally used to provide a more even surface, with enhanced frictional characteristics between itself and footwear which both contribute to increasing the speed of the game. However research has shown that synthetic surfaces compared to traditional grass surfaces can place the body under increased stresses. Various tests which have reported increased spinal shrinkage (Reilly and Borrie, 1992) and peak plantar pressures (Ford et al., 2006), demonstrate a potential detrimental aspect of the synthetic surfaces used in the modern game. This highlights an increasing need in the modern era of the sport to consider the effects of the surface on potential overuse injuries. Two main types of synthetic surfaces are currently used. These are commonly known amongst field hockey participants as sand-based and water-based Astroturfs. Sand-based surfaces use a pile fabric attached to a stable subsurface covered with sand which is held in place by the strands of pile fabric (Haas Jr., 1982). Water-based surfaces are similar in design with the sand replaced by water. Water-based surfaces are now the standard for elite hockey matches. The majority of club hockey is played on both types of surface with larger elite clubs using the more expensive water based pitches more frequently. Many club players will participate on various standards of surfaces with many of the sand based surfaces used at this level not suitably maintained or very worn out. This may lead to a loss in the cushioning properties of the surface. This makes it problematic to draw conclusions from a single surface type such as sand based, that can be considered valid for all sand based surfaces. Furthermore many players training and warm ups involve running on hard surfaces such as concrete. The effects of harder surfaces on kinetics linked to overuse injuries needs to be considered as a possible contributing factor to injury occurrence. Shoes designed specifically for road running may be more suited to activities on harder surfaces, than those designed for synthetic sports surfaces. However, it has been reported that owning multiple pairs of footwear to participate in a sport has been shown in a running cohort to increase the prevalence of injury by 50% (Walter et al., 1989). It would appear that when choosing the appropriate footwear a single design that copes best with the factors influencing injury may be a crucial strategy for injury prevention.

Choosing the correct footwear to reduce the potential for overuse injuries requires consideration of the duration and intensity of the sport in question to assess the causative factors (Eils *et al.*, 2004;Popovich *et al.*, 2000). Field hockey participation typically involves matches that are played over two 35 minute periods and often preceded by a thorough warm up, and typically two training sessions a week. The duration and intensity of field hockey participation helps maintain physical fitness which has a positive impact on people's lives (Wilmore, 2003;Mersy, 1991;Williams, 1997;Myers *et al.*, 1999;Fentem, 1978;Burnham, 1998;Hegde, 2003;Maxwell, 2004;Dugan, 2007). Injury is a factor that can reduce or even stop participation altogether and therefore has a negative impact on the health and well being of an individual. In field hockey this is a significant issue with 10% of amateur hockey players' participation reported as being restricted by injury (Roberts et al., 1995). Injuries can occur due to a single traumatic event such as a fall or a collision with a stick, ball or opponent or they can be an overuse injury. Field hockey involves high intensity locomotion activities which expose the musculoskeletal system to loading which can cause overuse injuries (Bennell *et al.*, 1996a;Korpelainen *et al.*, 2001;Bennell *et al.*, 2004;Schwellnus *et al.*, 1990). Overuse injuries occur due to repetitive stresses applied to an area of the musculoskeletal system (Stanish, 1984;Hreljac and Ferber, 2006;Bennell and Brukner, 2005). Whilst not sufficient to cause a fracture or tear with a single impact, over time repetitive stress can lead to degeneration and eventual failure of the tissue. As well as impact forces during locomotion, diet, muscle strength and flexibility have been identified as factors affecting the prevalence of such injuries (Fredericson *et al.*, 2006;Micheli, 1986;Rolf, 1995).

Overuse injuries were reported to make up 18.4% and 31.7% of injuries for male and female elite hockey players respectively, with 22% of injuries reported for female players being soreness of shins. Other common overuse injuries in the lower extremities include anterior compartment syndrome, stress fractures, Achilles tendinitis, plantar fasciitis, shin splints, and chondromalacia patellae (Ross, 1993). Interval training compared to running at a consistent speed was found to increase the likelihood of overuse injuries to the tibia (Wen et al., 1997). With intermittent speeds of locomotion reported in field hockey participation (Spencer *et al.*, 2004a;Spencer *et al.*, 2006) the nature of the sport exposes the participants to a particular risk of suffering an overuse injury. Recommended treatment for overuse injuries involves a period of rest and correction of intrinsic and extrinsic factors (Palsson and Karlsson, 1998). Knowing the causative factors of these injuries is needed if external influences such as footwear and surfaces are to be used where possible to reduce the prevalence of overuse injuries. The causative factors can be identified using kinetic and kinematic biomechanical analysis which allows the influence of footwear and surfaces to be measured. Such information can then be used to identify the best conditions to reduce the likelihood of a participant sustaining an overuse injury.

1.2 Aims and objectives

The overall aim of this thesis is to investigate footwear choice and surface factors influencing overuse injuries in field hockey participation. The thesis reviews, implements and adapts previous methodologies to identify kinetic and kinematic variables linked to overuse injuries. Testing various footwear designs mechanically and during human locomotion this study provides valuable information for choosing the correct footwear for reducing the likelihood of the occurrence of an overuse injury. The objectives of this investigation include:

- To review the kinematic and kinetic factors influencing overuse injury, and the influence of footwear choice on these factors (Chapter 2).
- To investigate the pressure distribution in footwear uppers; the alignment of the ground reaction force vector and tibia; and the relationship between ground reaction forces and acceleration of the tibia, during field hockey specific movements (Chapters 3-5).

- To investigate the effects of commercially available footwear on shock attenuation using drop tests, and various kinetic variables measured during human locomotion (Chapters 6-11).
- To identify if the methodologies used in this thesis can identify footwear that will reduce the rates of overuse injuries in a general population of field hockey participants. Furthermore, if this is possible identifying a system that can asses footwear practically in a large population of participants.

1.3 Rationale for the study

This section gives a brief rationale of the problems as well as the methodologies that will be used to achieve the objectives that are all contributory to the overall aim of this thesis.

Peak pressure measured under the plantar surface of the foot has previously been linked to overuse injuries in the feet (Ghani Zadeh Hesar *et al.*, 2009;Hennig and Milani, 1995;Freeman, 2002;Guldemond *et al.*, 2008). Research has also reported that a proportion of the forces acting through the footwear on the human body are applied through the uppers of the footwear and not the plantar surface (Hosein and Lord, 2000). Pressure is also applied to upper regions of the foot due to tight fitting shoes (Rudicel, 1994). The peak pressures applied through the uppers of footwear would appear to be the cause of corns and calluses which are common on the lateral side of the 5th metatarsal (Freeman, 2002). There is currently a paucity of research into the levels of peak pressures applied to the foot through the uppers of footwear. Research within this thesis will investigate the applied peak pressures between the uppers of the shoes and the feet during field hockey specific movements. By adapting an in-shoe pressure measuring device, pressure between the uppers of footwear can be measured during human movement. These results may help identify design considerations for field hockey shoes as well as a variety of other sports involving repetitive specific movements.

While multidirectional movements in hockey such as cutting may pose a potential injury to the lateral sides of the feet, they may also increase the risk of an overuse injury in the tibia compared to forward running. This is due to a movement strategy that may expose the participant to potentially damaging loads at a relatively large angle to the longitudinal axis of the tibia. Research has identified the alignment of the skeleton as one of the most important factors to consider when designing running shoes and Orthotic inserts (Nigg, 2001). When a bone experiences a force not acting along the axis of the bone (known as a bending force), the total stress on the surface of the bone can be multiple times larger than the same force acting along the long axis of the bone (Nigg and Herzog, 1999). Clinical and experimental investigations provide evidence that stress injuries occur at the site at which the maximum tensile stress due to bending occurs (Mizrahi *et al.*, 2000b;Daffner, 1984). The bone's ability to resist bending moments has been identified as a factor that can reduce the risk of a stress fracture occurring (Milgrom et al., 1989). These different abilities of bones to resist such bending between individuals may be a cause of variability between participants in sports to avoid such injuries. It is clear from the previous research that the alignment of the various bones in the human body compared to the applied force may be a factor in the onset of injury. The body's ability to orientate its lower limbs to align the applied resultant force along the long axis of the tibia could reduce the likelihood of developing an overuse injury. The movement of the body can be effected by the insole texture. Insole texture has been reported to influence the proprioception of the musculoskeletal system through the plantar region of the foot (Waddington and Adams, 2003). Socks with different textures made to change the sensory input at the plantar region, have also been shown to significantly change the pressure distributions under the feet (Chen et al., 1995). Research within this thesis will investigate the effects of different textured insoles on the alignment of the tibia and the resultant GRF vector. The results of this study will provide evidence of the effects of increased proprioception on the alignment of the lower extremities and the GRF vector. This may assist in selecting the correct insole within field hockey footwear.

While cutting movements may pose a larger potential for injury, the most common movement in field hockey is forward locomotion during walking, jogging and sprinting (Spencer *et al.*, 2004b;Spencer *et al.*, 2005). Due to their repetitive nature, these types of motion should be the major areas of concern for overuse injuries. During impact with the ground due to human locomotion, a transient shock is experienced through the musculoskeletal system, known as an impact shockwave or heel strike transient. Impact shock has been highlighted by many studies as a factor causing many overuse injuries (Snel *et al.*, 1985;Zhang *et al.*, 2008;Milner *et al.*, 2006;Verbitsky *et al.*, 1998). By attaching accelerometers directly to the bone through inserted pins and also to the skin, impact shock can to some extent be measured (Kim and Voloshin, 1992). Skin attachment methodologies have been found to increase the magnitude of the acceleration signal due to the skin artefact (Lafortune et al., 1995a). However through use of a low-pass filter the component of the signal due to skin artefact can be separated (Shorten and Winslow, 1992). During forward linear human locomotion various GRF variables have been linked to impact shock including vertical loading rate, peak braking force, and time to impact peak (Oakley and Pratt, 1988;Hennig and Lafortune, 1991;Laughton *et al.*, 2003;Bus, 2003). GRF data can be normalised between participants in studies by expressing GRF parameters in bodyweights (BW). Previous research reported a direct relationship between BW and peak GRF magnitudes (Frederick and Hagy, 1986). Evidence such as this has led to expressing GRF in BWs as a standard practice in published research. (Munro *et al.*, 1987;McClay *et al.*, 1994;De Wit *et al.*, 2000;Diop *et al.*, 2005;Seegmiller and McCaw, 2003). Normalising data for BWs allows effective comparisons between data from the same and different studies. Body mounted accelerometers are affected by the angular velocity of the shank, gravity, position of the accelerometer and its mounting (Nigg & Herzog, 1999) and any differences in one participant's body at the area of attachment. This makes it difficult to compare values between participants within the same study and even more so between studies, hence GRF data being a useful tool for facilitating comparative studies. The most effective use of force data to identify impact shock is unclear, with different variables measured in previous research. Using GRF variables calculated across many previous studies, the relationship between these variables and the acceleration signal measured through a shank mounted accelerometer are investigated within this thesis. The results of this research can be used in analysis of the influence of footwear on GRFs linked to overuse injuries.

Footwear has been identified as influencing kinetic and kinematic data by many previous studies (Stacoff *et al.*, 1991;Bates *et al.*, 1983;Waddington and Adams, 2000;Perry *et al.*, 2007;Nurse *et al.*, 2005;Menant *et al.*, 2008;Ki-Kwang *et al.*, 2005;Cheung and Ng, 2008;Bates *et al.*, 1993;Dufek *et al.*, 1991;Simeonov *et al.*, 2008;Stacoff *et al.*, 2001;Bishop *et al.*, 2006). As well as the influence footwear may have on kinematic and kinetic data collected in this thesis, previous research has identified speed (Perry and Lafortune, 1995;Burnfield *et al.*, 2004) and surfaces (Riley *et al.*, 2007;Riley *et al.*, 2008;Hardin *et al.*, 2004;Dixon *et al.*, 2000;Stiles and Dixon, 2006;Stussi *et al.*, 1997) as factors influencing characteristics of human movements linked to overuse injuries. The locomotive speed of a hockey participant may be difficult to control without affecting the performance, particularly during a match situation. As surface and footwear have been identified as affecting kinetic and kinematic data correct choice of these factors can therefore have a positive influence on overuse injury prevalence.

This thesis will investigate if certain designs of sports shoes can be recognised as most effective across a broad population in reducing the occurrence of overuse injuries, through analysis of kinetic variables linked to overuse injuries. This will be carried out by recording kinetic data during human locomotion with participants in a range of shoes typically worn by field hockey participants. However, firstly the footwear is assessed using a mechanical drop test to identify cushioning properties. Knowledge of the cushioning properties of the footwear allows the evaluation of the effects of cushioning on kinetic data measured during shod locomotion. The influence of footwear and speed of locomotion on localised plantar pressure is then investigated using an in-shoe sensor. Pressure data is recorded during shod locomotion at $3.3 m.s^{-1}$ (jogging) and $5.0 m.s^{-1}$ (running) on a synthetic sports surface in a biomechanics laboratory. By identifying specific regions of the non dominant foot, the influence of the footwear on localised loading of the plantar region of the foot at the two different speeds is identified. This provides peak pressure information that can be used to compare the footwear for injury related peak values. The effects of footwear can be compared between the two speeds to test whether the influence of the footwear has an effect on the recorded data in both conditions. Using the same methodology, in-shoe GRF characteristics are investigated between the footwear conditions. The same GRF impact characteristics that were compared to tibial shock earlier are investigated. Using the information collected earlier, factors that had been identified as linking most strongly to impact accelerations can be investigated allowing the evaluation of the footwear.

The final variable investigated will be impact shock. The same methodology used earlier to record accelerations of the tibia is employed to measure impact shock. The data is recorded during shod locomotion at $3.3 m.s^{-1}$ and $5.0 m.s^{-1}$ on a synthetic sports surface and on a concrete surface. The results of this study will provide information for factors influencing overuse injuries, comparing the effects of the footwear and the surface at the two different locomotion speeds. This information will allow increased knowledge for field hockey participants when choosing appropriate footwear. Through identifying pressure, GRF and tibial acceleration data linked to overuse injuries, the footwear is then compared across all the data presented to provide an overall evaluation of each footwear design. This will include the mechanical cushioning values to allow comparisons between mechanical and human locomotion based testing.

The overall results of this thesis will provide evidence of the influence of footwear designs on factors linked to overuse injuries as well as providing new methodologies to be further investigated. It is unclear if one ideal shoe can be identified for a sport such as field hockey. However, the findings of this thesis will identify how footwear can influence the potential for overuse injury for individuals, as well as across a population of participants.

1.4 Scope of the investigation

The boundaries of the scope of this investigation were:

- To focus on the influence of footwear on injury potential during field hockey participation, including playing in a match, training and warming up. This investigation is focussed on participants who prioritise injury prevention when making footwear choices. Therefore the influence of footwear on performance characteristics in terms of improving speed and agility will not be investigated in this thesis.
- To investigate the effects of commercially available footwear and not to design new footwear or adapt existing footwear.
- To use non-invasive methodologies in collecting kinetics and kinematics.

1.5 Need for this study

Due to the wide variety of footwear currently used by field hockey participants, there is a need to identify the effects of footwear on the potentially detrimental kinetic and kinematic factors relating to injury. Identification of footwear suitability to surface conditions may allow participants to reduce their risk of sustaining an overuse injury which would restrict participation.

1.6 Ethical approval

Appropriate ethical approval was sought, and granted by the University Ethics Committee[1]. All participants provided informed consent to participate in the various experimental studies.

1.7 Structure of this thesis

This thesis is set out over 12 chapters. Chapter 1 introduces the background and aims of the thesis, providing the reader with the reasons that a thesis study of this kind is needed and how the thesis investigates new areas of research. Chapter 2 consists of two structured reviews. The literature review firstly identifies injuries relating to field hockey participation and the kinematic and kinetic factors influencing these injuries. This forms the basis for the following section of the review, which investigates previous research identifying how footwear affects the kinetic and kinematic data. Due to the nature of the relationships between injury, kinetics, kinematics and footwear, there is a certain amount of cross over in the reviewing of previous research. Some key research studies are described in depth to provide valuable evidence of methodologies used to collect kinematic and kinetic data in order to successfully investigate the effects of footwear. Other factors affecting kinetics and kinematics in shod locomotion are then discussed. In

[1]The major part of this study was conducted at Staffordshire University with a small part also conducted at the University of Central Lancashire. Appropriate ethics committees at both of the universities approved the relevant sections of this study.

particular, surface effects are focussed on due to their considerable influence on kinetic and kinematic factors.

The literature review leads this research to a series of scientific studies, investigating new and previous methodologies for measuring kinetic and kinematic data in relation to overuse injuries (Chapters 3-5). The first investigation (Chapter 3) explores a newly developed technique, adapting a plantar pressure sensor to measure the distribution of the force through the lateral side of the uppers of a sports shoe. Various sports specific movements are tested, with peak pressures at specific anatomical points reported. The second methodological investigation (Chapter 4) records the angular difference between the long axis of the tibia and the resultant GRF vector during a cutting movement. Using a force plate and three dimensional motion tracking system to identify the angle of the GRF vector and the tibia respectively, the research investigates if the surface texture of an insole affects this alignment. The aim of this study was to identify whether the possible increased levels of proprioception affect the movement strategy of the lower extremities to place less stress on the musculoskeletal system. Chapter 5 studies the relationship between GRF variables and accelerations measured at the tibia with a force plate and a shank mounted accelerometer respectively. The shank mounted accelerometer's mounting and attachment techniques used throughout this thesis built on information from previous studies.

A software program was developed using Matlab (Mathworks Inc, USA) to calculate GRF and tibial acceleration variables. The results from this research identified correlations between GRF variables and the magnitude of the tibial acceleration, which has been linked to overuse stress injuries in the tibia. This study provides evidence of the suitability of skin mounted accelerometers and GRFs to measure the potential for overuse injury in footwear which is carried out later in this thesis.

In Chapter 6 a mechanical test of six pairs of commercially available footwear typically worn during field hockey participation is performed. The mechanical test involved dropping a weight from two known heights into the heel pad region of the shoes whilst they were secured to a force plate. GRF variables previously investigated in Chapter 5 were calculated using the software programme (Matlab, Mathworks Inc, USA) developed and used from the previous Chapter. This provided GRF characteristics for each footwear that would allow comparison with human locomotion tests later in this thesis. This in turn provides evidence of the validity of such a mechanical test to demonstrate footwear influence on the kinetics and kinematics linked to overuse injury during human movement.

Chapters 7 to 9 explore the effects of the footwear introduced in Chapter 5 on human locomotion. More specifically Chapters 7 and 8 use in-shoe plantar pressure sensors to investigate participants' kinetics during jogging ($3.3 m.s^{-1}$) and running ($5 m.s^{-1}$) on a synthetic sports surface. While describing the movements of the participants investigated in this thesis, from Chapter 7 onwards the terms 'jogging' and 'running' will be used to describe human forward locomotion at $3.3 m.s^{-1}$ and $5 m.s^{-1}$ respectively. This will allow easier to follow discussions of the various factors influencing the data reported. Chapter 8 investigates peak pressures at various locations under the plantar region of the foot in the various pairs of footwear. Comparisons of the variables are investigated between the groups of participants as well as the differences for each individual participant, for each footwear condition. Chapter 8 analyses the force applied to the plantar region of the foot using in-shoe pressure measurement sensors which have been found to be highly correlated to vertical force plate measurements. The research reports the various GRF variables that are also calculated in Chapters 5 and 6. Comparisons of the variables are investigated between the groups of participants as well as the differences for each individual participant, for each footwear condition.

Chapter 9 uses a skin mounted accelerometer to measure tibial accelerations during locomotion at a sports arena, on a hockey specific sports surface and also on a concrete surface. The same six pairs of footwear tested in the previous Chapters (6-8) with the addition of a new design of running shoe are examined. The participants ran and jogged at the same speeds as in Chapters 7 and 8 on both surfaces. Data recorded was compared between the groups of participants as well as the differences for each individual participant, for each footwear condition.

Chapter 10 provides a final summary and discussion of the findings for the research undertaken in this thesis. The effects of footwear choice and the practicalities of testing footwear on an individual basis for field hockey participants is presented, whilst providing recommendations for areas of further research.

Chapter Two

Literature Review

2 Literature review

The literature used in this review was identified through initial searches in the electronic databases PubMed, SportDiscus and ScienceDirect. Further literature was sought from the references within these publications. A narrative literature review methodology was employed. The MESH terms used were: ‘field hockey’, ‘sports’, injuries’, ‘running’, ‘ground reaction forces’, ‘plantar pressure’, ‘tibia acceleration’, ‘footwear’ and ‘stress fractures’.

2.1 Injury risk in field hockey participation

In this section of the literature review, the movements involved in field hockey and how they expose the participants to the risk of suffering an injury are investigated from previous research in field hockey and similar sports.

2.1.1 Locomotive characteristics of field hockey participants

The detrimental effects of the foot to ground impacts that lead to injuries during sports participation are related to how often an impact occurs and the duration of the activity (Eils *et al.*, 2004;Popovich *et al.*, 2000). There is a clear paucity of research focussing on the movement strategies of hockey players during competitive match situations and training at any level compared to other sports. A relatively small study investigating the movements of 14 international players throughout a match has been previously performed (Spencer et al., 2004b). The research identifies periods of locomotion but does not report directional characteristics of motion. From observations of field hockey matches most of the movements are forward linear movements. However, there are movements such as stopping, side stepping and cutting performed, particularly when in close proximity to the ball. The main focus for overuse injuries should be on the most common movements, therefore this thesis will concentrate on linear movements; however there will be some considerations of more complex multidirectional movements. The results shown in Figure 2.1 show the breakdown of the movement strategies which were concluded by the authors as being similar to soccer, rugby and Australian rules football. Furthermore it has been reported that anaerobic power of elite field hockey male participants is similar to soccer players, and higher than basketball (Reilly and Borrie, 1992). These sports are played on pitches of similar sizes and require balls to be moved between the team of 11 to 15 players in order to get the ball over a line in a certain manner. Therefore to be a successful team, players are required to perform short periods of fast top speeds with intermittent jogging and walking across these sports. Furthermore, the required endurance fitness levels of the participants needs to be at similar levels in order to compete over 70-90 minute matches. These similarities between field hockey and other sports are important to demonstrate. With the paucity of field hockey movement information published, useful data may be obtained from research investigating other sports.

Investigations into soccer have reported distances covered by elite players to be between 10 and 11km over 90 minutes (Stolen *et al.*, 2005;Bangsbo, 1994). If field hockey is considered to be played at the same intensity for a 70 minute match an estimation of distance covered would be between 7.8 and 8.6km. Over a 5km run the average athlete has been reported to experience approximately 3000 foot to ground impacts (Laughton et al., 2003). Therefore an estimation of the number of foot to ground impacts for a participant during a match would be approximately 4700 to 5100. If players' movement characteristics are such that they are exposed to excessive loading during participation, with this number of impacts over the amount of time participated, they are clearly exposed to the potential to suffer an overuse injury. Although the total distance covered in a match was not reported, researchers have investigated the movement intensity during field hockey matches (Figure 2.1). Considering the detrimental effects of increased speed (Perry and Lafortune, 1995;Burnfield *et al.*, 2004;Weyand *et al.*, 2000) and that striding and running intensity during a match combined with jogging make up 46.1% or 32 minutes of the match, these movements involving effectively a series of take-offs and landings should be the focus of research due to the increased exposure to a potential injury for the athlete (Keller et al., 1996). However, it should be recognised that by running at different speeds, specific loading of the musculoskeletal system will change. Therefore, specific sites at which repetitive stresses are experienced may change throughout a match, allowing for periods of recovery. In terms of peak plantar pressures there is clear evidence of this occurring. Research has found that while some areas of the foot experience increases in peak pressures, others experience a reduction as participants' speeds increase (Weyand et al., 2000). However, for tibial injuries it would appear that as the speed increases there may be an exponential increase in injury risk. This evidence is apparent in research which reported that interval training compared to running at a consistent speed, increased the likelihood of overuse injuries to the tibia (Wen et al., 1997). Although evidence from bone pins inserted at various sites have shown that certain areas of the tibia experience localised stresses with different movement strategies (Ekenman et al., 1998). Impact shockwaves travel up the length of the tibia at impact increasing in magnitude with speed(Perry and Lafortune, 1995). Therefore the application of this detrimental shockwave does not change so much its position, just its magnitude. This evidence may be why interval training that is similar to participation in hockey matches, places the tibia at increased risk of injury. This highlights the importance of understanding the breakdown of locomotion speeds during sports participation in general.

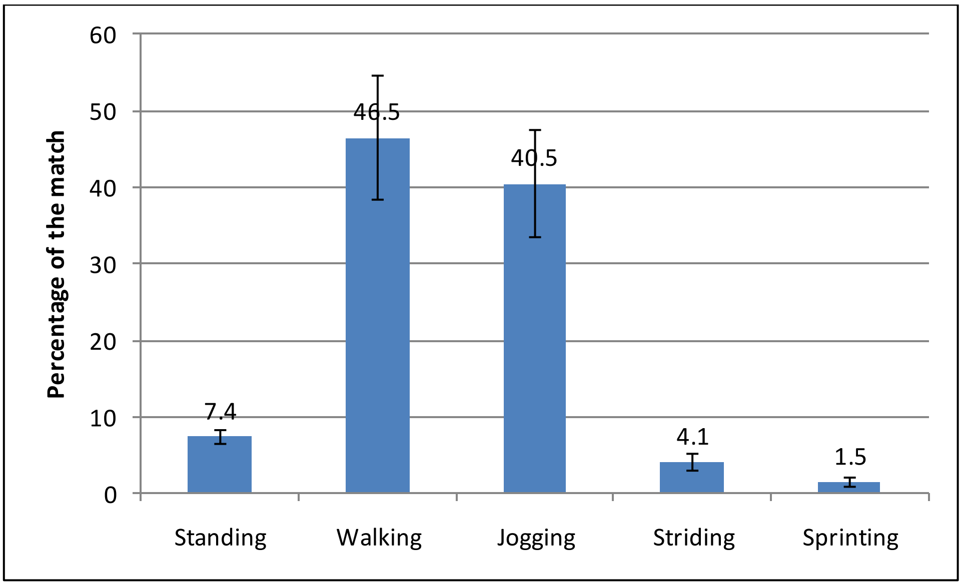

Figure 2.1 Mean movement strategy of 14 elite field hockey players as a percentage of the time each player spent in a single match adapted from Spencer et al. (2004b)

Further research by the same group of authors (Spencer et al., 2005) produced a more in-depth study recording 14 field hockey players' movement strategies over a four day tournament in which the players participated in three matches. The study added to the database of knowledge on the movement strategy of elite field hockey players during a single match as well as investigating the possible effects of fatigue on the athletes due to the high frequency of matches. However it must be acknowledged that this is a relatively small study and the movements in the matches could be influenced by the standard of the team being investigated as well as the fitness, ability and tactics used by the opposition. Furthermore, the participants in this match were elite level athletes and would therefore be physically fitter and play the match at a faster pace, covering further distances during the game than would be the case in a typical club match. However, currently this research offers the most information available from a scientific source and is therefore the best available scientific information on field hockey movements within elite hockey. The results reported from the study found a significant increase in the time spent standing (7.4±2.0, 11.2±2.7 and 15.6±5.6%, $P<0.05$) between matches 1, 2 and 3 respectively. This increase in time spent standing suggests an increase in player fatigue as may be expected. A significant reduction in the time spent jogging was also found between games 1 and 3 and games 1 and 2. Whilst it would appear there may be less demand on the shoes being worn in a fatigue affected match the effects of fatigue have been shown to influence kinetics and kinematics exposing an athlete to an increased risk of injury (Mizrahi *et al.*, 2000a;Tsai *et al.*, 2009;Mizrahi *et al.*, 2000c;Coventry *et al.*, 2006;Bisiaux and Moretto, 2008;Schlee *et al.*, 2006;Nummela *et al.*, 1996;Derrick *et al.*, 2002;Nagel *et al.*, 2008). Therefore the fatigued conditions may expose the participants to an increased risk of injury despite the reduction of the jogging and faster movements in a match.

If fatigue affects the kinetics and kinematics of players significantly as suggested in the literature, it may influence the effect the footwear has on a player's injury potential. This evidence suggests that it may be possible to reduce the chance of injury by changing shoes (between matches) that alter kinetic and kinematics more effectively. During fatigued conditions this may reduce the likelihood of injury occurrence in such intense competition formats. As a match progressed it may also be possible to change shoes as a player fatigues in order to reduce the exposure to detrimental loading characteristics. However, changing of shoes has been highlighted as being linked to an increase in injury prevalence (Walter et al., 1989) so the influence of any such recommendations require further investigation.

As well as considering the movement of hockey players within hockey matches, training and other sporting activity during a typical week could also influence injury occurrence. The breakdown of field hockey participants' specific movements and their intensity over a typical week, which may include several training sessions and fitness work, was not found to be reported in any literature. It may be the case that if there was such a study it would be very specific for individual clubs depending on training times, competitions and coaching methods used. Considering the approximation of foot to ground impacts made in this section, a typical week could involve over 30,000 such impacts during field hockey participation. Thus the hours of participation during a week that involve locomotive movements may be a factor that could be adjusted for players with an increased susceptibility to overuse injuries, with other sports such as cycling and swimming replacing such activities.

Since the time at which the studies investigating movement during a field hockey match (Spencer *et al.*, 2004b;Spencer *et al.*, 2005) and the data collected further in this research project were performed, a new self pass rule has been introduced. This rule came into effect at the start of the 2009/2010 season and allows players to play the ball immediately after an offence has occurred without having to pass to another player. From observations of field hockey matches from club level to international level this appears to have decreased the number of breaks in the game and may therefore have increased the distance players will move in each of the periods. This may expose the players to an increased chance of suffering an injury. However research on the movements of players under these new rules is not yet available. The previous data available should for the moment be considered as it is the closest data for guidance on locomotive movement within field hockey. There does need to be an acknowledgement of the potential for possible inaccuracies in the data due to the new rule changes and further research on the movements within a match needs to be carried out.

During field hockey participation holding a stick may affect the player's movement and thus should be a factor to consider when using previous research from other sports to identify injury potential in field hockey. There is little information on the effects of holding a stick during locomotive movements other than specific skill performance such as striking the ball (Bretigny et al., 2008). As there is an alteration in the movements, this may affect the stresses on the musculoskeletal system. It may also have an effect on the movement forces involved as having the controlled free movement of arms has been shown to enhance sporting performance in other sports movements such as jumping (Ashby and Heegaard, 2002). During human locomotion it was found that most kinetic and kinematic variables were highly correlated ($r>0.90$) when comparing with and without arms free to move (Umberger, 2008). During hockey matches and particularly in training when not involved in play with the ball many players will hold their stick in a single hand in such a way as to minimise its effects on their movement. This may be a factor to consider when investigating the protective function of field hockey shoes. However, whilst the previous research does suggest some small differences in locomotion, the effects appear to be relatively small and for the majority of movements performed in field hockey participation, may be further minimised through holding the stick in one hand. Therefore research involving movements with free arms will be considered suitable to investigate the effects of kinetic and kinematic data on injuries and footwear in this thesis.

2.1.2 Introduction to injuries associated with field hockey participation

Numerous studies have published information on sports injuries over many years (Tucker and Alexander, 1954;Quigley, 1959;Coughlin and Baker, 1965;Blonstein, 1974;Muckle, 1982). Over the years, increases in sports participation, the commercial value of many sports and the athletes involved have identified an increasing need to perform extensive research into sports specific injuries (Juma, 1998;Stevenson *et al.,* 2003). With a general paucity of research in the area of field hockey, other invasion sports-specific research can provide evidence of the injury potential of field hockey participation. As this study focuses on overuse injuries caused due to foot to ground impacts during locomotion, the distance covered, intensity and duration of movement during participation are important factors. Therefore using sports injury research from the sports identified as similar to field hockey provides further information on typical injuries which field hockey players are susceptible to.

2.1.3 Typical injuries

The results of a survey of physicians and trainers (n=42) who worked with football, baseball and basketball teams reported the top fifteen most common foot and ankle injuries (Figure 2.2) (Moseley and Chimenti, 1995). The results agree with other research that reported the most common injury from the data collected during sporting activity is ankle ligament injury (Beynnon et al., 2001). Such injuries appear to be the main area for concern across sports including field hockey and research has reported ankle sprain as the most common injury (Murtaugh, 2001). Furthermore, data reported by the Football Association suggested ankle sprains to be the most common injury found in soccer with 73% of ligament injuries occurring at the anterior talofibular and 14% occurring at the medial side (Woods et al., 2002).

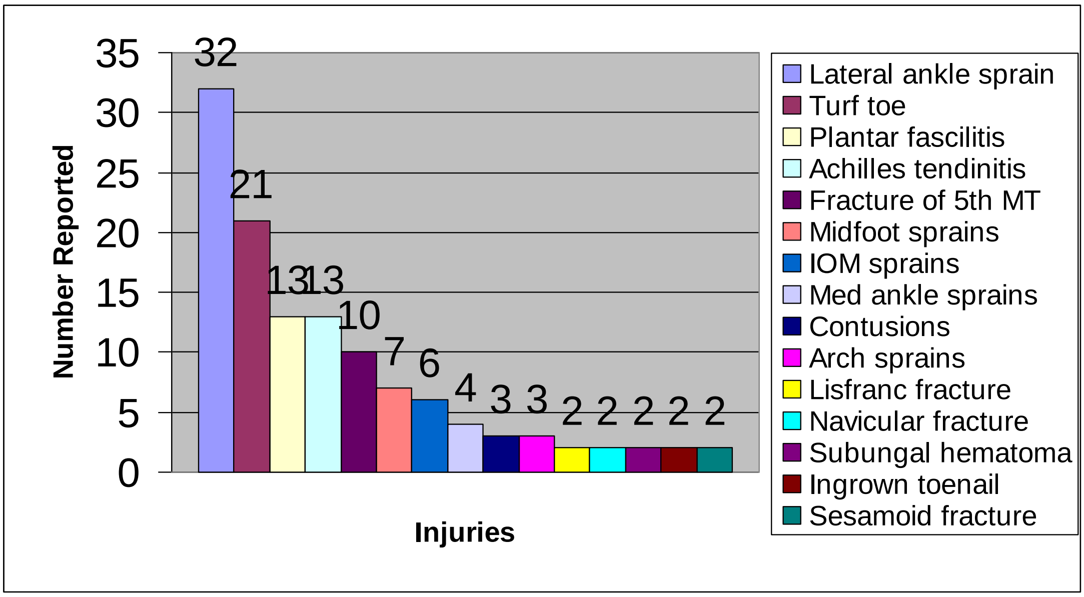

Figure 2.2 Fifteen most commonly reported foot and ankle injuries, adapted from Moseley & Chimenti (1995)

In the United States alone, more than two million individuals suffer from ankle ligament trauma each year with severe ligament sprains occurring in over half the cases (Beynnon et al., 2001). Footwear research has shown that shoe design can assist in the reduction of lower limb injury prevalence (Barnes and Smith, 1994). The sports reporting frequent ankle sprain occurrence all involved dynamic, multidirectional movements suggesting a link between specific injury frequencies and the movement characteristics of the sport in question. Similar kinematic and kinetic variables were found within the gait of participants who sustained inversion sprains. These included a longer total foot contact time, a higher loading underneath the medial and less underneath the lateral border of the foot and a medially directed pressure distribution at first metatarsal contact forefoot flat and heel off (Willems et al., 2005a). Therefore identifying participants of sports such as field hockey, who report kinetic and kinematic data linked to injuries, will allow for participants to be targeted and adaptations to their movements to be made. Footwear that could alter kinematics and kinetics may achieve this as well as reducing ankle sprains in a wider population.

Ankle sprains occur during a single excessive load about the ankle joint. In field hockey participation this can happen due to contact with another player or just simply stepping awkwardly maybe due to the circumstances of play such as reacting to a pass. These incidents which cause an ankle sprain are not very controllable and so the amount that can be done to limit such injuries is by its nature limited. Overuse injuries are caused by repetitive loading at a certain part of the musculoskeletal system, therefore if the loading can be identified and changes made, the occurrence of an overuse injury may be stopped.

Stress fractures, particularly in the tibia are a type of overuse injury associated with intense sporting activity (Ekenman *et al.*, 2001;Smrcina, 1991;Milgrom *et al.*, 2003;Mizrahi *et al.*, 2000a;Iwamoto and Takeda, 2003). A stress fracture occurs due to repetitive stresses applied to a bone, whilst not sufficient to cause a fracture with a single impact but over time leading to degeneration and eventual failure of the tissue. Footwear, surface, bone density, diet, muscle strength and flexibility have been identified as factors affecting the prevalence of such injuries (Fredericson *et al.*, 2006;Micheli, 1986;Rolf, 1995;Nattiv, 2000;Milgrom *et al.*, 1992). A stress fracture forms when a bone is remodelling due to a new loading environment. Inadequate bone tissue is replaced with tissue suited to withstand the new conditions. During this period the bone is vulnerable and any imbalance or excessive loading during this period may result in the occurrence of stress fracture symptoms (Bennell et al., 1996b). This process highlights the need for loading of the system in order to maintain and develop suitable bone structures for sports participation (Vico *et al.*, 2000;Bennell *et al.*, 1996a). However too much loading is detrimental hence stress fractures in the lower extremities occurring in populations who participate in high levels of sporting activity (Bennell *et al.*, 1996a;Korpelainen *et al.*, 2001;Bennell *et al.*, 2004;Sormaala *et al.*, 2006). Interval training used in field hockey participation has been specifically linked to stress fractures in the tibia (Wen et al., 1997). It has also been found that athletes with personality traits that make them more competitive and motivated and therefore less likely to stop due to discomfort as well as an overall higher level of training are a higher risk group (Ekenman et al., 2001). The tibia has been identified by many studies as being a common site for stress fractures (Rolf, 1995;Sasimontonkul *et al.*, 2007;Ekenman *et al.*, 1998;Milner *et al.*, 2006). In particular the anterior cortex of the tibia is highlighted as an area of concern (Boden and Osbahr, 2000). Bone deformation in the tibia increases up to eight times during locomotion compared to standing (Rolf et al., 1997). It is clear that stress fractures in the tibia are an area of concern for participants of field hockey and other sports. Another area of concern when considering stress fractures in athletes is the Calcaneum (Sormaala et al., 2006). Such injuries may be more prevalent than reported in the literature due to misdiagnosis because injuries to the calcaneum are often mistaken as an ankle sprain due to their close proximities (Gilheany, 2002). Furthermore, radiography has been found to not detect stress fractures of the calcaneum in the middle and anterior of the calcaneum (Sormaala et al., 2006), increasing the likelihood of a misdiagnosis. While diet, bone density and flexibility may assist in the body's ability to protect itself from suffering a stress fracture in the tibia and calcaneum, it is the magnitude of the impact shocks and the duration of physical activity that appear to be the main cause. By reducing the magnitude of impact shocks through footwear, surface and movement strategies a reduction in injury prevalence may be possible in a population. The durations of activities known to cause large impact shock transmissions such as running should also be controlled with adequate rest periods.

Metatarsals have also been identified as areas susceptible to stress fractures during field hockey and similar sports participation (Metz, 2005;Iwamoto and Takeda, 2003). During long distance running it was found that at the end of a run, plantar pressure loading patterns had altered so the heads of the metatarsal were under increased weight bearing (Nagel et al., 2008). Furthermore, a study has identified that if the first ray fails due to muscular fatigue, disease or trauma then the second ray in turn will become exposed to overloading increasing the possibility of injury to that site of the plantar surface (Jacob, 2001). It would appear that as an athlete fatigues they may be increasing their susceptibility to stress fractures in the foot that may not be apparent from investigating pressure under the plantar region of the shoes when an athlete is non-fatigued. Similarly it has also been shown that during running, as an athlete fatigues the body's ability to attenuate the impact shock leads to an increase in the accelerations measured at the tibia through a skin mounted accelerometer (Voloshin et al., 1998). This evidence suggests that in order to identify the potential for overuse injuries in the lower extremities a combination of GRF, tibial accelerations and plantar pressure can all provide evidence that could identify the causes of stress fractures. While Bone pins can offer further valuable information, due to the discomfort and invasive nature of this methodology it would not be a practical method across a large sporting population. Using non-invasive methods to collect kinetics and kinematics may be more suited and practical for assisting athletes who have a history of overuse injuries.

2.1.4 Preventing an overuse injury

Preventing an overuse injury from occurring is a major goal of sports scientists. This section investigates how previous training, duration and intensity of activities can be adapted to reduce the risk of overuse injuries.

While previous physical activity would seem to predispose individuals to develop a musculoskeletal system that was less likely to suffer an overuse injury due to conditioning, research has suggested this may not be the case. No correlation was found between army recruits who participated in sporting activities prior to training and those who did not with the occurrence of stress fractures (Swissa et al., 1989). Studies such as this have to be approached with caution as a participant who is physically fitter than another may have increased their body's ability to withstand larger forces leading to stress fractures. However due to this, the participant may expose their body to more intense exercise over a longer duration. In many studies this may be a problem that affects the findings. If an athlete is more physically fit and more protected from injury due to previous training, this may lead them to expose their body to more intense exercise and more volatile movement strategies. Furthermore, this would mean that using participation in an activity such as a field hockey match to compare participants suffering an injury may not be an effective way of identifying causative factors of injury. A thorough analysis of the differences between individual's movements during field hockey would be required to identify injury causing factors for individual participants.

It would appear that getting the balance between enough exposures to lower extremity impacts to promote health in the musculoskeletal system without reaching the point of injury may be the correct way to avoid stress fractures. However controlling an athlete's movement characteristics strictly is not necessarily possible in a sport such as field hockey where their personality, opposition and team mates will affect players movements and thus dictate the stresses to which their body is subjected in a competitive environment (Ekenman et al., 2001). Using army recruits who undergo controlled physical activity provides sports scientists with large amounts of comparable data. Using army recruits (n=1357), research has found that a week's rest in the middle of an eight week training course did not significantly reduce the incidence of overuse lower extremity injuries (Popovich et al., 2000). The results of this relatively large study of army recruits found that the lowest injury group was the one who ran the most miles with intermittent rests, recommending running and marching with single days of rest in between as having a positive effect on lower extremity overuse injuries such as stress fractures. This is an important finding for populations of athletes such as field hockey players identifying the strategy of intermittent days for training and match participation. Field hockey tournaments would appear to be set up ideally for this with day rests between games at the highest level.

2.1.4.1 **Returning from an overuse injury**

Preventing an injury from occurring in the first place is problematic with various types of injuries to consider. However, a previous injury diagnosed correctly allows strategies to prevent re-occurrence of the specific injury that the athlete is known to be susceptible to. One of the main issues with stress fracture re-occurrence is athletes resuming physically demanding sporting activities prematurely. A gradual resumption of levels of loading is recommended when returning from overuse stress fractures (Bennell and Brukner, 2005). Researchers using the term ‘sequence of prevention’ identified the factors that caused the injury, the severity of the injury, what could be done to reduce the injury risk in the future, and finally how effective these implementations in preventing future similar injuries were (Van Mechelen et al., 1992). Using the ‘sequence of prevention’ method with an overuse injury suffered in the lower extremities would therefore be as follows: Firstly identify the physical activity causing the overuse injury; followed by diagnosing the severity of the overuse injury; then reducing the risk through periods of rest; training involving less volatile impacts; reducing impact loading through footwear and surface changes;

and finally assessing the state of the injury after a suitable period. This process requires knowledge of the influence of footwear, surface and physical intensity and duration on injury risk. If these three factors can be fully understood, this method of reducing injury re-occurrence should see be effective.

Preventing injury re-occurrence may be easier to devise such a strategy for, as the type of injury is known. Trying to prevent the risk of all injuries is a more challenging area for applied sports biomechanics. However by identifying factors linked to overuse injuries such as physical activity intensity, movement strategies, footwear and surfaces, application across larger populations could reduce the occurrences of injuries in the first place. The following section investigates and defines what kinetic and kinematic variables can be identified as linked to overuse injuries.

2.2 Kinetic and kinematic data linked to injuries

Kinetic data investigated in this thesis includes GRFs measured in three dimensions from force plates, in-shoe reaction force and in-shoe pressure both measured using an in-shoe pressure measuring device. Kinematic data in this thesis describes the motion of the human body and is collected through a 3 D opto-electric 8 camera system. This section introduces the technology for collecting kinetic and kinematic data that shall be investigated in this thesis. How similar data has been collected in previous research is identified and discussed. Furthermore, the links between various characteristics measured and overuse injuries are investigated. This provides the basis for section 2.3 and 2.4 which investigate non-footwear and footwear effects respectively, on the data that is reported to be linked to injury.

2.2.1 Ground Reaction Force

During locomotion the human body exerts a force upon the ground which results in an equal and opposite force known as a Ground Reaction Force (GRF). GRFs experienced between the foot and the ground provide propulsion of an athlete in the desired direction. However during human locomotion, GRFs which have been found to be multiple in magnitudes compared to the body weight of the individual, expose the musculoskeletal system to potentially injury inducing forces (Hamill *et al.*, 1983;Kaplan and Heegaard, 2000;McClay *et al.*, 1994;Clarke *et al.*, 1983a;Frederick and Hagy, 1986;Munro *et al.*, 1987;Nilsson and Thorstensson, 1989;Collins and Whittle, 1989;Wiegerinck *et al.*, 2009;Keller *et al.*, 1996;Lees and Field, 1985).

2.2.1.1 **Measuring ground reaction forces**

Force plates (otherwise referred to as force platforms) have been used to collect force data during human locomotion since the mid 20th Century (Marks, 1953). For the past half a century force plates have been used to measure GRFs during sporting movements (Ramey, 1970;Ramey, 1972). Force plates require rigid mounting into the ground within laboratory or sports surfaces or on a raised walkway. Force plates provide three axis' of forces Fx, Fy, and Fz, three turning forces Mx, My, and Mz about the x, y and z axis' respectively and the location of the centre of pressure. When measuring GRFs there is often a need to normalise the data between participants due to different body masses by measuring impact forces with the ground in Body Weights (BW) (Mass x Acceleration due to Gravity). This is possible as previous research reported a direct relationship between BW and peak GRF magnitudes (Frederick and Hagy, 1986) and is a widely used methodology in published research (Munro *et al.*, 1987;McClay *et al.*, 1994;De Wit *et al.*, 2000;Diop *et al.*, 2005;Seegmiller and McCaw, 2003). Although this method is a useful tool in comparing the results of a wide range of participants it has the potential to negate the importance of the effects of body weight on the kinetics involved. A number of authors have reported increased GRF peaks as a function of increased load carriage (Kinoshita, 1985;Tilbury-Davis and Hooper, 1999). A study investigating force transmissions for different backpack loads applied during walking found there to be no significant differences in the peak GRFs recorded at initial ground contact (Holt et al., 2006). If the BW of a participant does not have an effect as much as is thought, it may be that a participant of small mass may report larger impact force characteristics when measured in BWs. Research has also highlighted the need for height variability to be considered when scaling to detect a difference in gait data between participants (Pierrynowski and Galea, 2001). This research agrees with other research which found that without any normalisation, height and weight accounted for 7-82% of the variance (Moisio et al., 2003). Within this thesis participants will be tested in multiple footwear allowing comparisons between the shoes for the same individual and thus the same height and weight.

Collecting natural human locomotive movement is an issue as participants are often too aware of the need for them to land their feet on a specific area of the ground. This effect known as force plate targeting may have an effect on human locomotive kinematic and kinetic variables. Previous research has investigated this potential problem (Challis, 2001). In the study the participants (n=7) were positioned to record the fourth footfall (normal position). Participants ran in these conditions striking the platform at 3.2 m.s-1 ±5%. A short condition was also investigated with the starting position moved forwards 50cm from the normal starting position, as well as a long condition where the starting position was moved 50cm back from the normal starting point. The participants were informed to strike the plates in the short and long conditions. During the normal conditions the participants reported no need to target the force plate and made successful contacts. No significant differences were reported in shear impulses and force plate contact time, suggesting no significant change in the ML and AP momentum and thus velocity. Significant differences were reported for the vertical GRF impact peak for short (1.52 BW), normal (1.66 BW) and long (1.93BW) conditions. Also significant differences were reported for the time to the impact peaks for short (14.7%), normal (14.8%) and long (12.5%) conditions. The increase in impact force may be due to the vertical velocity of the lower extremities at impact. If the step length was longer and the speed of the participants remained constant it suggests that the participants would have to propel their centre of mass to a higher height to remain in flight for a longer time period. As the centre of mass of the participant would be accelerated under gravity over a longer time periods it would suggest that the velocity at impact would therefore be larger. Research has found that a larger velocity is correlated to a large impact force peak (Elvin *et al.*, 2007a;Elvin *et al.*, 2007b). The kinematics recorded in the study by Challis (2001) reported significant differences ($P<0.05$) in the angles of foot, shank and thigh segments during locomotion when force plate targeting. The results of this important study highlight the need to ensure participants are running naturally and avoid force plate targeting. This may be an impossible factor to eradicate but every effort should be made to reduce the effects.

GRFs can be measured using in-shoe plantar pressure measurement systems and pressure mats. However these systems can only measure forces perpendicular to their surface and may not cover the whole area under the plantar region of the foot and have inherent inaccuracies when not used in conjunction with a force plate. Force platforms provide the standard form of collecting 3D force data. However restrictions regarding cost of platforms and embedding them into a surface as well as the effects of force plate targeting on human motion may restrict the accuracy and practical use of such devices.

An innovatively designed 3D force measurement system attaching sensors to the bottom of shoes was investigated by (Liedtke et al., 2007). The force measuring sensor on the soles of the shoes produced very little differences between results recorded from the force plate when investigating the vertical GRF (2.3% RMS difference). However there was a much larger difference when investigating the mediolateral (ML) (37.2% RMS) and anterior-posterior (AP) (10.1% RMS) force components. The Resultant GRF vector was found to have the smallest difference (2.2% RMS). The results suggest there are increased errors in the lower magnitude shear forces. This provides evidence that studies such as (Munro et al., 1987) which investigate lateral dynamic movements, would not be suitable as well as any possible stability issues caused by the increased height associated with thicker soled footwear.

However for investigation of the vertical and resultant GRFs in linear forward motion, the shoes would appear a practical alternative to other measurement devices.

2.2.1.2 **Ground Reaction Forces and their relationship to overuse injury**

During initial ground contact the human body is exposed to a large GRF reached in a very short time period referred to as an impact peak (Dayakidis and Boudolos, 2006;Hennig and Lafortune, 1991;Stuelcken and Sinclair, 2009;Gottschall and Kram, 2005;Nachbauer and Nigg, 1992;Nilsson and Thorstensson, 1989). Reaching this large peak force in a relatively short time period results in transient shocks travelling upwards through the musculoskeletal system which have been linked to various overuse injuries and can be measured by attaching accelerometers to the human body (Auvinet *et al.*, 2002a;Lafortune and Hennig, 1992;Voloshin *et al.*, 1998;Verbitsky *et al.*, 1998;Lafortune and Hennig, 1991;Lafortune *et al.*, 1995a;Hennig *et al.*, 1993;Mercer *et al.*, 2002;Hreljac, 2004;Zhang *et al.*, 2008;Auvinet *et al.*, 2002b). Research has found that in a study of 13 athletes with tibial stress fractures and 23 without, GRF characteristics did not significantly differ between the groups (Bennell et al., 2004). The authors suggest that GRF characteristics should not be used to identify athletes at risk of suffering stress fractures of the tibia. However a similar study of 20 athletes with a history of tibial stress fractures and 20 without, found significant differences between groups with higher loading rates and higher peak impact shock measurements recorded in the athletes with a history of tibial stress fractures (Milner et al., 2006). The two studies investigated similar GRF characteristics suggesting that group size and variability may have been a factor in providing significant values for the study by Milner et al (2006). However with the lack of significant differences found in the study by Bennell *et al* (2004), it suggests using GRF characteristics to identify stress fracture injury risks may not be a suitable practice. Using direct measurement of impact shock should therefore be considered a more effective way of comparing footwear influence on individuals, to assess the risk of suffering a stress fracture in the tibia.

As well as forward locomotion, sports including soccer, field hockey, tennis, rugby, netball and basketball involve the participants making sharp cutting movements to achieve a rapid change in the direction of movement. In order to perform a cutting movement, a lateral force must be applied to the ground through the foot to produce a GRF which propels the body in the desired direction. A study of GRFs experienced by basketball players (McClay et al., 1994) found that during a cutting movement the mean mediolateral (ML), anterior-posterior (AP) and vertical components of the resultant GRF vector were 1 BW, 1.1BW, and 2.3 BW respectively. This differed substantially compared to the components of the resultant GRF vector found during running which were 0.2 BW, 0.4 BW, 2.5BW, for the ML, AP and vertical components respectively. Furthermore, when investigating the results from a range of studies involving running, the mean force peak of the ML GRF component was reported to be 0.29 BW (Munro et al., 1987). During human locomotion in an anterior direction the forces are relatively low compared to much larger components experienced during movements with a lateral direction of motion component. Sports incorporating these movements have relatively high rates of ankle sprains. Excessive traction allowing for high ML forces may be a causative factor. Furthermore the alignment of the GRF resultant vector and the tibia may place the bone under considerable bending force. When a bone experiences a force not acting along the axis of the bone (known as a bending force), the total stress on the surface of the bone can be multiple times larger than the same force acting along the long axis of the bone (Nigg and Herzog, 1999). Clinical and experimental investigations provide evidence that stress injuries occur at the site at which the maximum tensile stress due to bending occurs (Mizrahi *et al.*, 2000b;Daffner, 1984). From observations of field hockey matches and training sessions, the number of multi-directional movements is relatively low compared to forward locomotion movements. While these movements may contribute to the occurrence of overuse injuries, initial research in field hockey should focus on the most common movements when investigating overuse injuries relating to impact forces.

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2.2.2 Impact shock

The term impact shock is used to describe a shockwave that is transmitted through the musculoskeletal system form the feet to the head.

2.2.2.1 **Measuring impact shock using accelerometers**

Transient Impact Shocks experienced through the musculoskeletal system during locomotion can be measured using accelerometers attached to the human body. Accelerometers measure acceleration by means of a small mass suspended by a stiff spring element. When the accelerometer is accelerated, the small mass is displaced exerting a small force against a sensing element. This results in a small electrical output being produced that is proportional to the acceleration acting on the element (Valiant, 1990). An accelerometer will measure the acceleration component in a single direction only. However there are tri-axial accelerometers available, which will measure the acceleration components in all directions using three uni-axial accelerometer components. The same results can therefore be obtained by simply using three single direction accelerometers mounted in the correct positions. Studies investigating the transient shock waves occurring at impact during human movement have used accelerometers mounted to the body to measure the accelerations of the musculoskeletal system during the initial foot to ground contact period (Hennig *et al.*, 1993;Snel *et al.*, 1985;Hennig and Lafortune, 1991;Lafortune *et al.*, 1993;Lafortune *et al.*, 1995b;Clarke *et al.*, 1985;Voloshin *et al.*, 1998;Shorten and Winslow, 1992;Pohl *et al.*, 2008).

When measuring segmental acceleration using an accelerometer the magnitude of the acceleration measured by the accelerometer is dependent on:

- Bone acceleration
- Mounting interaction
- Angular Motion
- Gravity (Nigg & Herzog, 1999)

Measuring the angular motion may not always be practical and furthermore when comparing subtle differences in footwear and other factors the amount of change of angular motion may be minimal and have very little effect on the accelerometer signal when comparing between impacts with the ground during locomotion. The mounting of the accelerometer is a factor that may have an effect on the data particularly between studies where different attachment systems are used and applied by different researchers. Within studies the same researcher should use the same method of attachment in order to limit the effect that mounting of the accelerometer will have on the data recorded between participants.

In order to accurately measure accelerations in the skeletal system, the accelerometer needs to be as rigidly attached to the bone as possible. Previous research showed that by sufficiently strapping a metal holder on the skin and attaching the accelerometer, shock wave amplitudes can be accurately measured (Kim and Voloshin, 1992). However, later research found that peak axial accelerations recorded from skin mounted accelerometers were more than twice the magnitude of those registered at the bone through invasive direct attachment means (Lafortune et al., 1995a). Nigg and Herzog (1999) suggest the method of mounting accelerometers by screwing a pin into the bone will produce more reliable and accurate results as the pin connecting the bone and accelerometer will be rigid. The investigation by (Lafortune and Hennig, 1991) considered all four elements mentioned earlier that affect the magnitude of acceleration recorded by the accelerometer. By mounting the accelerometer via a pin through the tibia, then calculating the angular motion of the tibia about the ankle joint and finally considering gravity, they left only the acceleration through the tibia due to impact shock remaining from the resultant acceleration measured by the accelerometer. However previous research has been able to demonstrate it was possible to separate the two signal components (the signal component from the impact shock and the signal component due to resonance of the accelerometer mounting on the skin) through a frequency analysis and the use of a low pass filter (Shorten and Winslow, 1992). The risk of inaccuracy in the use of skin mounted accelerometers to collect skeletal impact shock data, can be reduced through effective marker placement, skin stretching techniques and the use of lightweight, rigidly attached accelerometers. Using the skin mounted accelerometer methodology has since been used in research with the accelerometer attached tightly to the skin at the anterior medial aspect of the tibia (Coventry *et al.*, 2006;Flynn *et* *al.*, 2004;Pohl *et al.*, 2008;Laughton *et al.*, 2003;Milner *et al.*, 2006). This position provides minimal skin interaction and minimal effects of acceleration due to the angular motion of the tibia about the ankle joint. Although the methods investigating impact shock using invasive bone mounted accelerometers and measuring kinematics would appear to provide the most accurate results. They require surgical procedures and expensive 3D measurement systems. A single axial skin mounted accelerometer system allows data to be collected more efficiently and without surgical intervention. Therefore, for analysing footwear in teams of hockey participants, would be a more favourable methodology.

Studies have however recorded impact shock further up the musculoskeletal system. One such study used an accelerometer attached by a belt at the lumbar region (Auvinet et al., 2002b). The accelerometer data was recorded at a relatively low frequency (100Hz) with a 50Hz low pass filter applied. The data presented reported peak vertical accelerations much lower than research measuring impact shock at the tibia. The accelerations reported were done so at a relatively high running speed compared to previous studies measuring tibial accelerations. Due to the attenuation of impact forces from the feet to the head in the human musculoskeletal system (Light et al., 1980) the magnitude of accelerations would be expected to be lower at the lumbar region compared to the tibia as it appeared was the case. Another factor influencing the peak accelerations would be the lower frequency used and the lower level set for the low pass filter which according to previous studies may have lost a proportion of high frequency impact acceleration producing lower acceleration values. The study provides further evidence that the position of the accelerometer on the body and the attachments influence the output signal and therefore make it difficult to compare between studies and even between participants within the same study. It is clear from the research reviewed in this section that accelerometers can be used to give a direct estimate of the acceleration experienced through the tibia, however due to the nature of the errors built into the signal and the differences in individuals' skin thickness or bone structure, comparisons between participants is questionable. GRF measurements can be normalised between participants and may offer a better methodology for comparing the effects of interventions such as footwear between individuals. Furthermore with accelerometers defining a value for impact shock that could be considered likely to cause injury may be too influenced by the data collection methodology whereas GRF could offer consistency that could be used more effectively by sports scientists. The relationship between GRFs and impact shock measure by accelerometers therefore needs to be investigated.

2.2.2.2 **Impact shock related to ground reaction force characteristics**

A study using an invasive bone mounted interaction for the connection of the accelerometer by Hennig and colleagues found that the peak tibial acceleration recorded by the bone mounted accelerometer occurred prior to the first GRF peak (approx 5ms) (Hennig and Lafortune, 1991). They conclude that there is not a simple relationship between the GRFs and the peak accelerations experienced. The study did report only a moderate correlation between the peak tibial acceleration and the peak first GRF peak. However there is a high negative correlation (r=-0.89) reported between the time to the first GRF peak and the magnitude of the peak tibial acceleration which supports the high correlation (r=0.87) also reported between the loading rate to the first peak and peak tibial accelerations. A shank-mounted accelerometer, a force plate and a 3D opto-electric motion analysis system were used to collect tibial accelerations, GRFs and kinematics of natural heel strike runners (n=15) performing their normal running style and forefoot running style (Figure 2.3) with and without foot orthotic devices (Laughton et al., 2003). Comparing the accelerometer data and the GRF data, the study reported significant ($P<0.05$) correlations between average loading rate and peak tibial accelerations for both forefoot ($r=0.70$) and heel strike ($r=0.47$) conditions. Interestingly the study reports stronger significant ($P<0.05$) correlations between instantaneous loading rate and peak tibial accelerations for both forefoot ($r=0.73$) and heel strike ($r=0.70$) conditions. This data is important for GRF data studies as it demonstrates a better correlation between tibial peak positive acceleration and instantaneous loading rates compared to average loading rates. It also suggests that GRF characteristics can predict tibial accelerations in forefoot running more effectively than in heel strike movement strategy. Other studies have investigated the loading rates between data points calculating the loading rate across 1ms of time (Bus, 2003), which will be referred to as the instantaneous loading rate. While average loading rates require identification of the 1st peak or impact peak which can vary due to the style of runner, the instantaneous loading rate does not require any such identification and therefore should be more reliable across studies as there is less human input into the analysis and therefore less chance of discrepancies between the results.

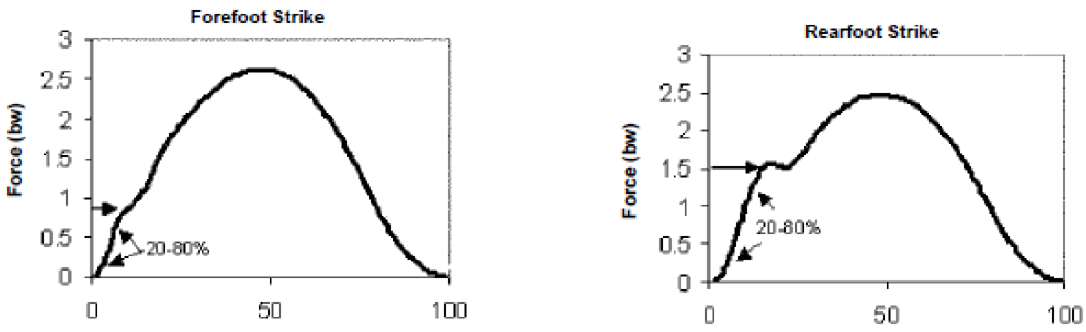


Figure 2.3 GRF time graphs of typical fore-foot strikers compared to a typical rear-foot striker. Adapted from (Laughton et al., 2003)

2.2.2.3 **Impact shock related to injury**

Impact shock experienced during human locomotive movement has been highlighted by previous research as a factor causing many overuse injuries (Snel *et al.*, 1985;Zhang *et al.*, 2008;Milner *et al.*, 2006;Verbitsky *et al.*, 1998). Excessive impact shock measurements have been recognised as being linked specifically to tibial stress fractures. An investigation analysing 40 athletes half with, and half without a history of tibial stress fractures was undertaken. It was found that the group with a history of stress fractures produced significantly higher impact shock measurements from skin mounted accelerometers (Milner et al., 2006). Further evidence of this link can be found in an almost identical study (n=60) (Pohl et al., 2008). The results found that larger accelerations were recorded by the athletes who had previously suffered tibial stress fractures. It should also be recognised that the transient shocks experienced through the human musculoskeletal system are essential for maintaining a suitable level of bone density to cope with the environmental and work load factors of an individual during locomotion. A lack of such impacts has been researched in space flights which have identified a significant reduction of bone density in the weight bearing bones during flights demonstrating the effects of extreme reduced impacts (Vico et al., 2000). It is unclear from the literature how much bone density plays a role in stress fractures due to impact shock. Lower levels of bone density have been linked to increases in the occurrence of tibial stress fractures in athletes (Myburgh et al., 1990). However, research investigating the links between stress fractures and bone density has found no significant links between the two measurements (Bennell *et al.*, 2004;Carbon *et al.*, 1990). In earlier research led by Bennell and colleagues investigating the same links in males and females there were no significant links reported between stress fractures and bone density in males (n=58). However, in females (n=53) lower bone density was found to be significantly linked to the occurrence of stress fracture (Bennell et al., 1996a). It would appear that in some populations bone density plays less of a role in the occurrence of overuse injuries. However the fact that some studies have found links to bone density and stress fractures suggests that in a wider general population increasing bone density will reduce the occurrence of stress fractures. Therefore activities involving non excessive impact shock should be considered valuable in developing the musculoskeletal system to resist injury. Reducing the impact shock through attenuating systems is therefore advantageous in reducing the occurrence of overuse injuries.

2.2.2.4 **Impact shock attenuation**

Research suggested that the body would attenuate impact shock relative to its magnitude in order to keep accelerations at the head constant (Hamill et al., 1995). This was found to be achieved by the body altering its kinematics to attenuate more of the transient shocks in the lower extremities, torso and neck before reaching the head. There are two types of shock attenuation mechanisms that can be considered, passive and active mechanisms. Both types contribute to the human musculoskeletal system's ability to attenuate potentially damaging levels of impact shock. Passive systems refer to systems such as the heel pad under the foot, cartilage and synovial fluid and muscles. While active systems refer to movement strategies, such as increased knee flexion, have been found to have a significant effect on peak impact shock experienced through the skeletal system. By flexing the knees during ground contact, the body is decelerated over a longer time period reducing the rate of loading which causes the impact shock (Lafortune et al., 1996). Muscles have also been highlighted as a major contributor to the absorption of shock due to their ability to transfer the kinetic energy into heat energy (Derrick et al., 1998). However, once cushioning properties have bottomed out below the calcaneum, the heel experiences a rapid upwards acceleration to stop the body from going through the surface. In shod conditions, the cushioning would be the heel pad tissue and the various materials used to cushion the impact in the rearfoot of the footwear, barefoot conditions would just be the heel pad tissue. To investigate the effects of a small amount of cushioning material under the heel pad, participants in a study ran in stockinettes (Oakley and Pratt, 1988). Three different insoles were tested in the stockinettes as well as a condition with no insole inserted in the stockinettes. The 3 insoles inserted were made of a cellular urethane foam, cleron and viscolas which had durometer 00 readings of 49, 63 and 49 respectively. Participants ran at controlled speeds between 3.3 and 3.6m.s^{-1}, in heel strike and forefoot landing styles, in the four stockinette conditions. Tibial accelerations were recorded using an accelerometer mounted using a bite bar, and GRFs using an embedded force plate. The data comparing the three materials, only reported a significant reduction in the tibial acceleration during a forefoot landing for the cellular urethane foam compared to barefoot conditions. Significant reductions in the loading rates were reported for the cellular urethane foam sole and the cleron compared to barefoot running. These results show that even a thin layer (6mm) of a material can significantly affect GRF characteristics and its effects on the human musculoskeletal system. However the results of the study also demonstrate that differences in foot position at landing can affect loading and impact shock experienced through the musculoskeletal system.

When the human body senses it is being exposed to larger impact shocks due to large high frequency GRF components acting through the plantar region of the foot, it has been shown that the human body will adapt its running style (De Wit *et al.*, 2000;Hennig *et al.*, 1996). Furthermore, various studies have identified that by purposefully adapting the running style of participants, the loading characteristics can be significantly changed (Oakley and Pratt, 1988;Laughton *et al.*, 2003;Lieberman *et al.*, 2010). The effects of such running strategies on impact loading of the musculoskeletal system are discussed later (2.3.2). Running style changes and adaptations of movements due to changes in cushioning of surfaces and footwear, demonstrate that large amounts of cushioning may not necessarily reduce detrimental impact kinetics.

2.2.3 Plantar pressure

When a ground reaction force is applied to the human body, the force applied to the plantar region of the foot is not done so uniformly across the surface of application. This leads to areas of relatively high and low localised pressure. An example of this is demonstrated in Figure 2.4 where the areas of red represent areas of high pressure while the blue areas represent areas of low pressure.

2.2.3.1 **Measuring plantar pressure**

Using a pressure measuring system (inserted inside the sports shoe) localised plantar pressure can be calculated over specific areas of the foot such as the metatarsal heads, mid foot or heel. These systems provide valuable data describing the distribution of the ground reaction force transmission through the cushioning in footwear to the plantar surface of the foot. Pressure plates (also known as pressure mats) are also able to measure pressure but over a larger area using similar technology as the inserts. During barefoot analysis they are an effective way of measuring plantar pressure, however for studies investigating localised pressure applied to the plantar region they are problematic as they measure pressure between the ground and the outer sole of the shoe. As shoe inserts directly measure the pressure applied to the plantar region of the foot, for investigations measuring the loading on the feet in shod conditions, they are used in many studies investigating such kinetics (Burnfield et al., 2004). Furthermore, testing of the F-scan in-shoe system that is used within studies in this thesis, it was found that they provide reliable measurements for peak pressures in shod testing (Ahroni et al., 1998). However it has been reported that the F-scan system has a significant effect on kinematic data. Increases in stride frequency and decrease in stride length were observed when wearing the system (Kong and De Heer, 2009). These results suggest there may be a decrease in the proprioception or an increase of slipping of the foot in the shoe, both leading to less stability and thus a reduction in stride length. However

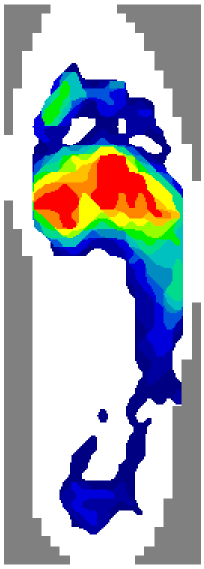


Figure 2.4 Typical plantar pressure recorded at mid-stance during running.

it should be noted that the differences are in the worst case, less than 3%. Furthermore they did not affect trends in all the kinematics analysed that were found with the increases in speed in the study. This would suggest that while such inserts will affect kinematics, if they are present in all the footwear being tested for all participants a reasonably fair comparison can be made.

2.2.3.2 **Plantar pressure related to injury**

Considering the general loading of the body during the foot to ground contact phase there is a need to investigate localised pressure at the plantar surface of the foot. Areas of intense localised pressure can lead to overuse injuries such as stress fractures of the metatarsal bones in the foot (Hennig and Milani, 1995). Corns and calluses which develop due to hyperkeratosis caused by excessive pressure (Freeman, 2002), have also been identified as occurring due to peak pressures measured at the plantar region of the foot (Guldemond et al., 2008). High localised plantar pressure has also been found to have a detrimental effect on blood flow increasing the risk of ulceration (Santos et al., 2003), Furthermore, peak pressures can also lead to general skin breakdown causing discomfort (Kelly *et al.*, 2000;Mueller *et al.*, 2005). All these injury outcomes can lead to a reduction in sports performance and participation. In particular risk groups such as diabetics, corns and calluses developed from peak pressures can lead to amputation or even fatality. Footwear and orthotics can be used to alter the distribution of peak pressures under the feet providing protection from the detrimental effects identified in this section.

2.2.4 Kinematics

Measuring kinematics during human movement can provide valuable information regarding the effects of factors such as surface (Riley *et al.*, 2007;Riley *et al.*, 2008;Hardin *et al.*, 2004;Dixon *et al.*, 2000;Stiles and Dixon, 2006) and footwear constructions (Morio *et al.*, 2009;McNair and Marshall, 1994). There are various methodologies for collecting kinematics to identify links between human movements and injury causing kinetics.

2.2.4.1 **Motion capture**

Video cameras and motion analysis digitising software provide a relatively cheap method of recording data that has been used in scientific research (McNair and Marshall, 1994;Yu and Hay, 1996;Auvinet *et al.*, 2002b). Video footage recording devices can record data at high frequencies of over 1000Hz. Such devices only measure in two dimensions and rely on a human operator identifying anatomical points and/or using tracking software that may have inaccuracies in identifying anatomical points under skin and clothing. Video systems have been used very effectively to synchronise with other systems and record events such as foot to ground contact and take-off (Auvinet et al., 2002b), and other sports specific movements (Mori et al., 2002). For this purpose they may be more accurate than 3D motion analysis as it is clearer when contact with the ground is in a frame of data. Various three-dimensional motion capture systems are currently available which provide accurate high frequency data through various methods. Opto-electric systems use infra red light cameras to track reflective markers attached at anatomical points on the human body. However research has found skin movement artefact can lead to substantial errors when investigating bone motion (Karlsson and Tranberg, 1999). Further research identified errors of 16, 5 and 3mm for the hip, knee and ankle joints, the study found that errors were strongly related to the amount of soft tissue present (Taylor et al., 2005). The use of these systems needs to be approached with caution. Smaller errors at areas such as the foot with less overlying soft tissue compared to the thigh segment where large amounts of soft tissue are present, will increase the error. This means that the ankle joint would provide more accurate data than the knee joint, although the complexity of the ankle joint provides its own problems. It would appear that selection of participants with less body fat and musculature may assist in accuracy of such data although selection of participants in this way may not reflect the overall population.

Another issue with using an opto-electric analysis system is that it requires the attachment of small reflective markers that may fall off participants, particularly during dynamic movements. Markers may also not be seen by the minimum of two cameras required due to clothing or limbs blocking the cameras' view. Furthermore, the system will not work effectively or at all outside during sunlight hours due to being flooded with infra-red light from the natural ambient light. There is also a human element to the identification of anatomical landmarks which the ease of identification can vary between participants in studies. The ability to palpate an area to identify sites for markers such as the anterior superior iliac spine can be relatively easy on people with a low body fat percentage becoming increasingly difficult as the participant's fat percentage increases, which also increases the movement of the skin during motion over a given anatomical point. Many participants may also not be comfortable in wearing the limited amounts of clothing required to allow markers to be attached directly to the skin and be visible. Tight fitting suits to which markers can be attached have been designed in an attempt to deal with such issues. However this may increase the errors of the system due to the movement of the material over the skin and also any problems it may cause when identifying anatomical points. A multi camera system is essential for successful data collection of many sporting movements. The more cameras used, the larger the capture volume can be and the less likely a marker will be hidden from less than two cameras during capture which will result in gaps in the data. Due to the new technology, support and cost of cameras, these types of systems are relatively expensive for a multi camera system. However for an indoor environment with sufficient room they offer an accurate way of measuring kinematics of human movement which has been used in many research papers (Donoghue *et al.*, 2008b;Louw *et al.*, 2006;Lloyd *et al.*, 2000;Arampatzis *et al.*, 2005;Chin *et al.*, 2009;Donoghue *et al.*, 2008a). Other systems recording 3D segmental motion of the human body such as Coda (Charnwood Dynamics Ltd) use active markers that omit an infra red signal. Although they do not have the problems with daylight as in opto-electric systems, the active markers are much larger and thus may have a detrimental influence on recording natural movements of participants. However, they are an effective alternative and used in many research publications (Menant *et al.*, 2009;Maynard *et al.*, 2003;Monaghan *et al.*, 2007). One of the reasons for their popularity is that they are more portable than many systems. Although not used in this thesis, such a system may be vital if a practical and portable integrated kinetic and kinematic analysis system to assess footwear is to be developed.

The reliability of 3D lower extremity kinematics recording the same movements on the same day and separate days was investigated by Ferber and colleagues, (Ferber et al., 2002). A number of participants' (n=20) lower kinematics were recorded during running on a 25m runway at 3.65 $m.s^{-1}$ by a 6-camera 3D motion analysis system (Vicon, Oxford Metrics, UK) recording at 120Hz. The motion analysis system used reflective markers at certain anatomical positions which were applied for each participant data collection period by the same tester. GRF data was recorded by a force plate embedded into the runway measuring GRF data at 960Hz. The participants returned a week later and the same data collection procedure was carried out. Five good trials of data were selected for analysis from each participant during each data collection day. Data was normalised to compare events over 100% of the stance phase and then data was compared. GRF values were reported to be more reliable than kinematic data. Overall, between day Intra-class correlation coefficients were less than that within same day data collection. The research suggests that for best practice, data collection involving kinetic and in particular kinematic analysis (due to increased variability between days) should be collected on the same day and ideally in the same session. By using the same tester to apply the reflective markers to anatomical points, the reliability between participant's data should be maintained as much as possible. The use of an experienced biomechanist or suitable guidance from manufacturers would also help restrict any variability due to inconsistencies with the marker placements. This will restrict the error in kinematic data between participants more than will already be present due to reasonable human error and participant anatomical differences. While there are different available systems for collection of kinematics, marker tracking systems provide the most efficient and accurate method. However due to the costs of such systems, availability of the systems to researchers may be the overriding factor in choosing which system to use.

2.2.4.2 **Kinematics related to injury**

Differences in recorded Kinematics have been shown to influence potential injury causing impact forces (Gerritsen *et al.*, 1995;Bishop *et al.*, 2006;Derrick *et al.*, 2002;Nigg and Segesser, 1986). According to a Physical Stress Theory the body's movement strategy is the most important factor that a physical therapist can use to adapt the stresses applied to the human body during motion, resulting in tissue adaptation which may lead to injury (Mueller and Maluf, 2002). Adjusting the heel strike movement characteristics have been found to influence loading rates at the hip joint (Bergmann et al., 1995). The single participant investigation by Bergmann and colleagues investigated running barefoot by a participant informed to run normally where the mean peak instantaneous loading rate reported was 68.0BW.s^{-1}; softly where the loading rate reduced only slightly to 64.9BW.s^{-1}; and with a hard heel strike where a relatively large (60%) increase in the loading rate was experienced reporting an average peak of 108.9BW.s^{-1}. This highlights the human body's ability to consciously adapt its running style to directly influence Kinetics experienced through locomotive movement. Valuable data that can help explain how footwear does not always influence kinetic variables as might be expected. The effects of footwear on such kinematics are discussed later in this thesis.

Kinematics provide evidence of detrimental rearfoot movements, such as excessively pronated and supinated feet and the rate of change in rearfoot angle, which have been identified as causing lower extremity injuries (Konradsen and Voigt, 2002;Ghani Zadeh Hesar *et al.*, 2009). There is however some conflicting evidence, as excessively pronated and supinated feet have been found not to have an effect on the occurrence of ankle sprain (Dahle et al., 1991). However the same study found athletes with the excessive foot conditions were more susceptible to knee pain. It would appear that in general an excessively supinated or pronated foot at landing which would suggest less stability, does expose the body to an increased risk of lower extremity injury. Footwear choice and other factors can assist in reducing these damaging movements identified through kinematic analysis. The various kinetic and kinematic factors linked to injuries in this section can all be influenced by a variety of factors which will be discussed in the following sections (2.3 and 2.4).

2.3 Non-footwear factors affecting kinetic and kinematic data

This section highlights speed, running strategies, inclined and declined locomotion and treadmill locomotion as key factors that may affect the outcome of kinetic and kinematic variables. These factors need to be investigated and considered as they may influence the biomechanical data collected and may also be used in collaboration with footwear to decrease potentially injury causing kinetic and kinematic factors.

2.3.1 Locomotion speed

Locomotion speed influences various biomechanical factors. Increases in rearfoot angles, impact forces, impact loading rates and impact tibial shock accelerations have all been reported as being significantly increased ($P<0.05$) during running compared to walking (Perry and Lafortune, 1995;Burnfield *et al.*, 2004). Changes in the speed at which a human being moves have also been reported as having a significant effect on peak pressures experienced on the plantar region of the foot, with even moderate increases in walking speed (0.95 $m.s^{-1}$, 1.33$m.s^{-1}$, and 1.62 $m.s^{-1}$) reported to produce significant differences in peak pressures (Burnfield et al., 2004). Results from a study by Taylor et al, (2004) also showed significant increases in localised peak pressure when the velocity of the participants increased. For plantar pressure studies all the evidence makes it clear that the speed of participants needs to be controlled if comparisons between footwear affects are to be correctly investigated.

To increase the velocity of human locomotion a participant needs to either increase their stride frequency or increase their stride length. To achieve any change in movement strategy, the forces applied to the ground to produce propulsive GRFs need to be altered. It has been hypothesised that greater human movement speeds are attained through increases in GRFs and not through increasing leg movement velocities which would lead to an increase in stride frequency (Weyand et al., 2000).

The study by Weyand and colleagues reported GRF data from a force plate mounted into a treadmill. Participants ran at various intervals from 3m.s^{-1} until they were not able to increase their speed any further. Average vertical GRF values were determined from the ratio of step time to contact time. Top speeds were recorded and they were placed into 3 categories (slow, average, and fast). Significant positive relationships were reported between top speed of runners and stride frequency (r^2 = 0.30) and also top speed and stride length (r^2 = 0.78), they reported similar results for ground contact time (r^2 = 0.30). The average forces were found to also increase as a function of top speed (r^2 = 0.39). The results show that runners able to achieve higher top speeds experienced larger average forces more frequently. This suggests a possibility of a higher injury potential although it should be recognised that as the runners will have been running with this gait pattern for sustained periods of time their body's ability to attenuate the applied stresses should have been increased due to increases in bone density. When considering the effects of footwear characteristic on GRFs there may be a need to test the top speed of athletes to recognise any potential injury risks associated with the various kinetic and kinematic differences experienced by runners of various top speeds. Due to the effects of locomotion speed on kinetics there is a need to compare footwear in conditions where the participants are moving at a known, consistent speed. Within this thesis, running and jogging at 5m.s^{-1} and 3.3m.s^{-1} respectively will be the controlled speeds used when comparing footwear. This will allow fair comparisons between footwear as well as the effects of speed to be investigated. Furthermore as the population will all be healthy active adults with no physical injuries or disabilities 5m.s.-1 will not exceed their top speed.

2.3.2 Running Strategies

Many studies have shown that during forward locomotive movement most athletes run with a heel strike style of running (Gerritsen *et al.*, 1995;Lieberman *et al.*, 2010;De Wit *et al.*, 2000;Dixon *et al.*, 2000). It is this heel strike that produces relatively large impact peaks and high loading rates. Therefore a style of running that could allow effective movement without, or with a restricted heel strike may be beneficial in reducing potential injury causing GRF characteristics.

By landing in a more forefoot manner the individual can increase the time over which the heel is accelerated upwards and thus reduce the average acceleration. To investigate this theory kinetics recorded during human locomotive movement of participants (n=18) performing heel strike movement characteristics and forefoot strike characteristics were recorded (Oakley and Pratt, 1988). The participants ran, in a stockinette with and without an insole in, using three different insoles. Tibial accelerations were recorded using an accelerometer mounted using a bite bar, and GRFs using an embedded force plate. Peak accelerations were significantly ($P<0.05$) reduced by 32.9% and the loading rate (which from the literature would appear to be the average loading rate up to the impact force peak) reduced significantly ($P<0.001$) by 86.8% from 3.19BW.ms^{-1} to 0.42BW.ms^{-1}. The results from this study show large changes in loading and accelerations experienced between two landing strategies. Research between habitual barefoot-forefoot running participants compared to shod-rearfoot running participants, has reported similar loading rates for the barefoot-forefoot runners (Lieberman *et al.*, 2010). Furthermore, barefoot-rearfoot running reported significantly larger rates of loading than habitual barefoot-forefoot and shod-rearfoot running. Further research investigating the effects of running style was undertaken by Arendse and colleagues. In their study they investigated the effects of three different running styles (Arendse et al., 2004). The styles were defined as:

- Heel Strike – Initial ground contact made with the heel followed by the midfoot and anterior proportion of the foot.
- Midfoot - Initial ground contact made with the midfoot with no heel contact during the gait cycle.
- Pose – A much more complex running style requiring 7.5 hours of training for each participant. The participants had to be able to run while aligning the acronium, the greater trochanter, and lateral malleolus while leaning forward and allowing the body to fall forward to initiate movement. At initiation of the movement, they had to lift the supporting foot via knee flexion, avoiding pushing away from the ground. They also had contact with the balls of the feet and maintained a flexed knee throughout.

The idea of the pose running style was to provide some of the force attenuation benefits associated with running backwards while being more practical. Various kinematic variables were reported as being significant between the pose running style and the other two more conventional styles. With stride length being the most notable with values of 2.20, 2.17 and 1.48m for the heel strike, midfoot and pose styles respectively. This smaller stride length may be linked to the significantly smaller vertical displacements of the sacrum and the heel in the pose running style. As research has highlighted the link between vertical velocity at foot to ground contact and higher force peaks (Gerritsen *et al.*, 1995;Zadpoor *et al.*, 2007), an increase in potentially detrimental vertical GRFs would be expected. Vertical impact force and loading rate after 25ms and at force peak were reported as being significantly higher in the heel-toe running style compared to the pose and midfoot running styles which produced similar magnitudes for vertical GRF at these times. It is not necessarily clear how accurate the results regarding the impact force peaks were as some midfoot strikers do not always produce a definite single impact force peak. The reliability of using the vertical force after $25m.s^{-1}$ again is questionable as the mechanics of the motion at this point in time may be very different, leading to less conventional loading patterns. Taking the instantaneous loading rate of force as reported by Hennig and colleagues across the ground contact phase may be a more realistic and fairer comparison across the running styles (Hennig et al., 1996). This would allow identification of when the magnitude of the increase in load was at its peak which may occur well before or after 25ms. The evidence for the studies by Oakley and Pratt, Liebermann and Arendse and colleagues suggest that landing with a forefoot strike is an effective way of reducing loading and impact shock (Oakley and Pratt, 1988;Lieberman *et al.*, 2010;Arendse *et al.*, 2004). Although the research would suggest that a forefoot landing strategy is a better strategy for reducing the magnitude of impact shock, there is conflicting evidence. Research investigating runners performing forefoot and heel strike landings in shod conditions were reported as experiencing significantly larger impact shocks during the forefoot landing style (Laughton *et al.*, 2003). This research did not report any significant differences in the vertical loading rates. The increase in impact shock may have been caused due to an awkward landing due to lack of training for the forefoot running style. A significant difference was reported in the Anteroposterior GRF loading rate. Magnitudes of 26.17 ± 8.72 and 9.46 ± 3.39 $BW.s^{-1}$ were reported for the forefoot and rearfoot striking strategies respectively. This would suggest that there is a quicker excessive breaking which may explain some of the differences between barefoot and shod conditions. In shod conditions it would appear that the evidence is conflicting regarding the benefits of forefoot running. The amount of training would appear to be the main differences between the studies. The participants in the study by Laughton *et al* (2003) were instructed to land on the ball of their feet and given a few trial runs to practice this. As mentioned, the training for the study by Arendse *et al* (2004) involved more extensive training of participants, to adapt their running style, while the study by Liebermann (2010) used runners who habitually trained in forefoot-barefoot running conditions. This may account for the differences between the studies and suggest that extensive training may be required in order to benefit from any adaptations in running style. However, what is clear from the results of these studies is that by altering a participant's movement strategy the GRF characteristics can be significantly changed both positively and negatively, thus affecting their susceptibility to injury. In terms of field hockey, explosive dynamic movements may be more difficult to adapt as optimum speed of movement would be a desired characteristic. However as reported earlier a good proportion of a match situation (40.5%) is spent jogging at elite level (Spencer et al., 2004b) and may be even higher at club and recreational level. Therefore adaptations to running styles during these time periods especially with players more susceptible to injury could be beneficial in terms of reduction of the prevalence of overuse injuries.

2.3.3 Dominant and non-dominant sides

Peak pressures reported in the forefoot found no significant differences ($P<0.05$) across a large amount of patients between left and right forefoot peak plantar pressure values (Guldemond et al., 2007a). Munro and colleagues reported only slight differences in vertical and AP GRFs, however a distinct difference was found between some participants' ML GRF impulse (Munro et al., 1987). In a study of soccer players significant differences were reported between preferred and non-preferred in landing, sidestepping and cutting, with the dominant foot experiencing higher pressure in propulsion and the non-dominant foot during landing (Wong et al., 2007). From the findings of these studies it would appear that during normal forward moving locomotion there is little difference between dominant and non dominant lower extremities. However during more dynamic movements involving changes in direction the dominant side should be a consideration. Therefore data collected on single sides of the human body within this thesis will use either the dominant or non-dominant side for all participants to reduce any influence that side dominance may have on results.

2.3.4 Treadmill locomotion

Human Locomotion data has been previously obtained through analysing human locomotive motion during treadmill running (McNair and Marshall, 1994;Voloshin *et al.*, 1998). While a treadmill provides an effective way of analysing human motion in a laboratory setting it is unclear if the motion being observed is similar to the natural over ground motion that will occur during sports participation. It is also important to recognise that treadmills themselves have rubber runways and may attenuate extra impact shock through their construction. As mentioned previously this change in surface compared to specific sports surface can have a significant effect on the GRFs experienced during human locomotive movement (Steele and Milburn, 1988;de Koning *et al.*, 1997;Gerritsen *et al.*, 1995). Therefore within this thesis, data will be collected in over ground locomotion only to gain more realistic data to that which would be measured during field hockey participation.

2.3.5 Age

In a study of sports participation injuries leading to a hospital visit, it was reported that 50% of the patients were aged between 10 and 19 years (Sherker and Cassell, 1998) and peak cases were also found in the 16 to 19 years group in further similar research (DeHaven and Lintner, 1986). These results may well be due to relatively high participation levels in the age group and not necessarily due to any increased physiological risks that may be related to the ageing. One of the underlying factors to consider with age related to injury is the negative effect age has been found to have on tissue adaptation (Hsu *et al.*, 1998;Mueller and Maluf, 2002;Scopacasa *et al.*, 2002). Bone density has been found to decrease with age (Nordin et al., 2008) starting as early as 30 years, with the rate of decrease increasing with age particularly after 70 years (Scopacasa et al., 2002). This reduction in adaptation and bone development restricts the ability of tissue to develop and sustain tolerance to stresses caused by field hockey play. However the health benefits of exercise have been reported as particularly valuable for people as they get older (Dugan, 2007;Tanaka *et al.*, 2004;Larson and Bruce, 1987;Karinkanta *et al.*, 2009). Therefore it could be argued that factors reducing sports and exercise participation in elderly groups should be an important area for health and longevity in general populations.

As well as age playing a role in the body's ability to resist injury there is evidence that age has an effect on kinematic and kinetics. A study investigated GRF and kinematic differences in self selected running speed and controlled running speed across a group of older (n=16, 55-65years) and younger (n=13, 20-35years) well-trained male distance runners (Bus, 2003). At the self selected speed the two groups showed significant differences (P<0.001) in their average speeds of $3.77 m.s^{-1}$ and $3.34 m.s^{-1}$ for the younger and older groups respectively. Across both the self selected and controlled speeds the older group produced significantly (P<0.001) shorter stride lengths and higher stride frequencies. Interestingly the vertical impact speed was significantly higher in the older runners during the controlled speed, but no significant difference was reported at the self selected speeds. As reported earlier, vertical velocity at heel strike increases the magnitude of the impact GRF peak (Gerritsen *et al.*, 1995;Zadpoor and Nikooyan, 2006). This suggests that the older group while running at the controlled running speed of $3.3 m.s^{-1}$ was being exposed to larger detrimental vertical impact forces and could be a contributory reason for the older group employing a slower self selected speed. It may well be that the older group selected their self selected running speed to a level that their body's judged to be comfortable. The initial loading rates for the older group ($106.9 BW.s^{-1}$) were similar to the younger group ($102.3 BW.s^{-1}$) for the self selected speeds, however the initial loading rates for the older group ($107.5 BW.s^{-1}$) were significantly greater (P<0.01) than the younger group ($85.5 BW.s^{-1}$) in the controlled speed conditions. This pattern was also reported for the peak impact forces where the peak forces were similar at the self selected speed for the older (1.89BW) and younger (1.89BW) groups, but the older group experienced significantly higher impact peaks (1.91BW) compared to the younger group (1.70BW) during the controlled running conditions. This suggests that athletes when given a choice will select to run at speeds that expose the body to similar forces across the age range. However when forced to run at a faster speed the younger athletes will be able to employ movement strategies to more effectively reduce themselves to GRF characteristics that have been linked to overuse injuries. The reasons that the older group may not be able to attenuate detrimental kinetics effectively may be due to the reduced range of movement at the knee and ankle joints in an older group compared to a younger group of athletes running (Bus, 2003). These results highlight that older groups are particularly at risk at higher velocities and furthermore controlled speeds. Within kinetic and kinematic data collection this needs to be considered when deciding on speed of motion to be analysed. In field hockey about 5% of the game is spent moving faster than a jogging pace (Spencer *et al.*, 2004b;Spencer *et al.*, 2005). This would therefore expose older players to larger detrimental GRFs for a significant part of the game. Footwear education needs to be targeted at those with greatest risk; the balance between protection from injury and performance of athletes with a larger susceptibility to injury due to factors such as age should focus on protection to promote continued participation. By reducing extreme detrimental kinetics, elderly populations can maintain physical activity which has been shown to increase balance and bone strength, but must be continued to maintain these benefits (Karinkanta et al., 2009).

In this thesis the age range will be controlled for testing of footwear for participants between 18 and 30 years of age. This is to reduce the variability in data that may be caused by the affects of age during locomotion at non-self-selected speeds.

2.3.6 Gender

Field hockey rules are the same for both male and female participants and many competitions are set up for the mixed gender form of the game which usually requires a minimum of five outfield players from each gender to make up a team of eleven players. However injury factors have already been identified as being different between the genders (Bennell *et al.*, 1996a;Beynnon *et al.*, 2001;Piasecki *et al.*, 2003) In general female athletes have been found to have greater susceptibility to various injuries compared to their male counterparts (Hewett *et al.*, 2005;Hennig, 2001). Furthermore, differences in foot shape and size have also been identified (Wunderlich and Cavanagh, 2001). With these differences, it is important to identify the effects of gender on collection of kinetic and kinematic data related to injuries.

Volleyball specific movements have been shown to produce different lower extremity kinematics between genders (Salci et al., 2004). Similar results were also apparent in a study of the lower extremities while performing sidestepping manoeuvres by (McLean et al., 2005). Significant differences ($P<0.05$) have also been reported in knee and ankle range of movement, contact position and peak angular velocities in a drop test from a 60cm height onto a force platform (Decker et al., 2003). While these results may offer further evidence that genders should be considered differently it should be highlighted that the average mass of the genders differed by 21.7kg (males=81.8kg and female 70.1kg) so a fairer test may have been to compare genders of the same body mass although this may provide its own problems as there may be significant physical differences from gender groups of the same mass. These could include height, body fat mass and muscle mass which may affect movement strategies therefore because of non weight factors. It would appear that gender should be considered when identifying injury potential and gender specific field hockey footwear may assist in reducing injury prevalence if the difference between the genders can be successfully identified. Furthermore within this thesis, when testing between pairs of footwear the participants used will be all male. This gender was chosen due to more accessibility of participants and footwear.

2.3.7 Body weight

Mass has been shown to be a significant factor in the occurrence of injury (Doyle and George, 2004). Obese groups have been found to have a significantly larger foot plantar surface area and also demonstrated higher peak pressures (Gravante et al., 2003). However results from the same study show that there were no significant differences between the percentage pressure distributions for obese and normal participants. For this thesis the participants will all be healthy participants who by way of a BMI check will not be allowed to take part if they fall into the obese category.

The BW of an individual may to some extent be controllable, although the ability of the individual to control this factor may vary greatly. Footwear may provide one of the most effective ways of altering the kinetic and kinematic data recorded during human locomotion in field hockey participation. Footwear is a factor that is very controllable for an individual. Therefore by investigating the influence of footwear on such kinematic and kinetic data, the effects of footwear on characteristics linked to injury may allow field hockey participants to make informed choices that immediately reduce their chance of suffering an injury. The next section investigates in depth the effects of footwear on kinetics and kinematics reported from previous research.

2.4 Effects of footwear on kinematics and kinetics

This section investigates the influence that footwear choice can have on the Kinematic and Kinetic data linked to injury. The section also considers the effects of surface as it is the combination of the surface and the footwear that influence many of the kinetic and kinematic factors influencing injuries. Some techniques for data collection are explained in this section for comparisons between previously collected experimental data.

2.4.1 Introduction to sports footwear

The development of footwear designs and technologies over the past century has led to many sports specific footwear becoming available. Wide varieties of materials and design concepts have been developed by shoe companies (McNair and Marshall, 1994). The construction of sports footwear has many factors to consider including cushioning, traction, mass, comfort, and upper design. A review of sports footwear reported that there appears to be no definitive design of footwear that will decrease the overuse injury potential, for all individuals across a population (Kersting and Bruggemann, 2006). However there may be certain designs that have a positive effect in general across a population (Barnes and Smith, 1994). The challenge therefore for sports footwear from this evidence, is to design footwear that would be effective in reduction of overuse injuries in a large population of sports participants.

Footwear used during field hockey participation has changed from soccer specific footwear with deep cleats designed to cut through natural turf, to rubber out soled shoes designed specifically for modern synthetic surfaces (Figure 2.5 a and b). In addition to field hockey specific footwear, players can be regularly observed at various levels wearing sports shoes designed for general sports use or specifically for other sports. Running shoes, soccer synthetic surface shoes, fell-trail running shoes, and soccer hard surface moulded boots (Figure 2.5 c to f) are used at club hockey level. Previous Olympic tournaments have seen top level players preferring running style shoes as opposed to field hockey specific shoes (Frederick, 2008). Running shoes are designed for protection during running but may not perform as effectively in terms of sporting performance as hockey specific shoes due to their grip characteristics. Furthermore they may have a detrimental effect on stability during lateral movements in particular, and foot protection from impacts from sticks and balls. These possible performance and protection characteristics need to be investigated to identify if there are footwear designs that are best used for field hockey. With players using running shoes it would appear to suggest that players are willing to potentially decrease their performance level in order to increase their perceived level of protection from overuse injury. In a review of the semantics across a variety of general footwear (n=36), footwear considered to be sportive were grouped in the same axis as dynamic, for young people, and popular (Alcántara et al., 2005). The designs and aesthetics of the footwear would appear to be a factor for players when choosing what footwear to purchase and wear. This may be one of the reasons field hockey players are observed wearing shoes not designed specifically for

a)Soccer natural turf shoes

b)Field hockey specific synthetic surface shoes

c)Running shoes

d)Soccer synthetic surface shoes

e)Fell/trial running shoes

f)Hard surface soccer moulded shoes

Figure 2.5 Typical shoes used during field hockey participation.

the sport. Other reasons for this phenomenon of wearing non hockey specific shoes may be due to cost, availability, perception of protection, and actual performance characteristics. Field hockey footwear in recent years has shown a move towards a more modern looking lightweight design, with seemingly less protection from ball impacts incorporated into the borders of the shoes. A paucity of evidence exists for the effects of these designs on kinetic and kinematic data linked to overuse injuries. There is a need for research to investigate the various design influences in order to provide information on injury protection relating to footwear choice for field hockey participants.

2.4.1.1 **Introduction to the influence of footwear on kinetics and kinematics**

It is reported in previous studies that footwear can have a significant effect on the GRF characteristics experienced during human locomotive movement (Clarke *et al.*, 1983b;Clarke *et al.*, 1983a;Nigg and Morlock, 1987). Footwear has been recognised as having the potential to reduce detrimental forces applied to the human body during locomotion (Mueller and Maluf, 2002). However it is not simply a case of shoes providing large amounts of cushioning. Footwear designs influence human locomotion kinetics and kinematics. This can result in shoes producing lower impact peaks in material tests whilst producing higher impact peaks during human locomotion, when compared to another shoe design (Aguinaldo and Mahar, 2003). Evidence of this adjustment of movement strategy when less cushioning is available was identified through a larger plantar flexion recorded during barefoot conditions compared to shod (Aguinaldo and Mahar, 2003). This demonstrates the body's tendency to adjust its movement strategy when the amount of cushioning available is changed. This movement strategy adjustment can also be seen in changes in surface characteristics where the body adjusts its leg stiffness between hard and soft surfaces (Ferris et al., 1999). The human body is clearly able to adjust its movement strategy when exposed to potentially detrimental impact loading. The amount of movement strategy adjustment may be difficult to directly relate to aspects of footwear design due to the complexity of the mechanics involved in human locomotion. The evidence across this research demonstrates that footwear cushioning does not reduce the detrimental kinetic data recorded due to the adaption of the human movements measured and reported in kinematic data. This highlights the problem faced by footwear designers in producing footwear that can help protect the human musculoskeletal system from injury. With the adaptation to less cushioning by individuals, designing field hockey footwear that reduces detrimental loading of the musculoskeletal system across a large population may not be possible.

2.4.2 Footwear effects on kinematics

A decrease in initial force peak has been observed due to the adjustment of the foot's kinematics (Hennig et al., 1996). Barefoot to Shod conditions in running were reported to produce a more forefoot landing strategy in barefoot running (Figure 2.6), resulting in a reduced first impact peak (De Wit et al., 2000). However this impact peak is reached in a much quicker time producing a higher loading rate which has been found to correlate more to impact skeletal shock.

Differences between shoe constructions have been found not to be significant when measuring lower limb influence kinematics (Bishop *et al.*, 2006). Bishop and colleagues reported an average increase of 12 degrees of dorsiflexion during running

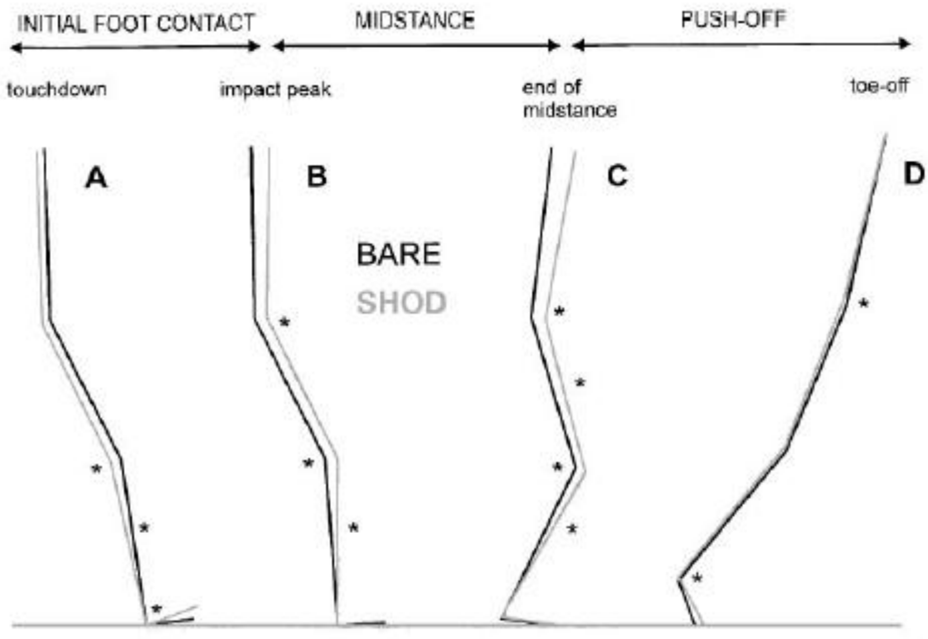


Figure 2.6 Kinematics of shod and barefoot running in the same population, adapted from De Wit et al, (2000). *=significant difference (P<0.05) between conditions

in both high and low cost running shoes compared to barefoot. The angle of the knee at impact has also been identified as influencing the magnitude of impact forces experienced during human locomotive movement (Nigg and Segesser, 1986) as well as landing from a jump (Elvin et al., 2007b). Increased knee flexion has been reported in barefoot compared to shod suggesting that it may be a coping mechanism of the body in reduced footwear cushioning conditions (De Wit et al., 2000). The evidence in this section appears to suggest that the recorded kinematics are influenced in a consistent manner in barefoot compared to shod conditions. However in general, shod conditions do not differ significantly between footwear designs across a general population. The lack of identification of significantly different variables across a population may be due to how individuals adjust to different footwear characteristics. Research investigating different midsole hardness concluded that kinetic and kinematic factors were largely dependent on the individual (Kersting and Bruggemann, 2006).

The effects of surface have been reported to show similar kinetic characteristics, with highest force peaks reported on the more cushioned surfaces, and highest loading rate values reported on the harder surfaces (Stiles and Dixon, 2007). In order to investigate this phenomenon, researchers constructed a 66.5kg model of the lower extremities incorporating previous muscular research to create a realistic model that could investigate the impact forces during ground contact phase of human motion without the adjustment of kinematics reported in actual human motion (Gerritsen et al., 1995). The study found that impact force peaks according to their model were largely influenced by plantar flexion reporting an 85N per degree in foot angle from measurements between 7.6 and 12.1 degrees of plantar flexion.

Recording Kinematics is useful for identifying why damaging kinetics related to injury are occurring, and can be related to damaging kinetics. However when testing currently available footwear effects on overuse injury during field hockey participation, the levels of impact loading, impact shock and peak pressures are the most important factors to consider. Therefore measuring these variables directly which can be directly related to overuse injuries is a more effective method for identification of the influence of footwear choice on overuse injury risk.

2.4.3 Footwear effects on impact forces and impact shock

Cushioning properties of shoes can be changed and are an obvious consideration when investigating how footwear protects a participant from impact forces and impact shock. Impact force and impact shock have been shown in material tests to be reduced with soft materials cushioning the impact (McNair and Marshall, 1994;Aguinaldo and Mahar, 2003)**.** Footwear cushioning may appear to be a reasonably straightforward concept in that reducing peak forces applied to an object by the use of soft material interventions increasing the time over which the foot is accelerated upwards during foot to ground impact. Various soft cushioning materials such as PVA, Gel, and Air are used to provide cushioning. However regarding human movement, simply using the softest materials may not be the best way to reduce GRFs that expose athletes to a higher prevalence of injury. In a presentation entitled "The Myth of Running Shoe Cushioning" the conundrum of how to define cushioning and also what is effective cushioning is highlighted (Shorten, 2002). With multiple ways of analysing footwear cushioning properties there is a need to review previous research to consider how the conclusions drawn and the results presented can be related to the identification of effective field hockey footwear.

The effects that various footwear conditions can have on GRFs and accelerations in mechanical and human impact tests have been previously investigated. A study by McNair and Marshall, compared four different shoes with various sole designs and shock attenuating materials (McNair and Marshall, 1994). The manufacturers designed the shoes with potential shock attenuation properties listed below:

- Shoe A – Double density EVA with a cantilever outsole;

- Shoe B – Double density EVA;
- Shoe C – Double density EVA containing air filled chambers;
- Shoe D – Encapsulated double density EVA. (McNair and Marshall, 1994)

The shoes were mechanically tested by dropping an accelerometer attached to a 9kg weight 5cm (approximate velocity at impact 1m.s^{-1}) into the shoe heel pad area. Human data was also recorded for the same footwear on a treadmill at 3.5m.s^{-1} with a 20g shank mounted accelerometer attached to the lower medial aspect of the tibia. The material test produced significantly different peak accelerations between all the shoes. The lowest peak acceleration was recorded in Shoe A (approx 9.6g), then Shoe B (approx 10.5g), followed by Shoe C (approx 10.7g) and finally Shoe D (approx 11.5g). No significant differences were found in the human running impact shock testing between the footwear, with accelerations in all footwear recorded at approximately 10g. In another study using mechanical and human running tests to investigate the effects of cushioning, three shoes from different manufacturers were compared (Aguinaldo and Mahar, 2003):

- Shoe 1 – Running shoe incorporating four cushioning columns made of multi-cellular urethane elastomer (Nike);
- Shoe2 – Running shoe incorporating four cushioning columns made of thermoplastic polyester moulded into a hollow, bumper-like unit (Iso-Dynamics);
- Shoe 3 – Running shoe with a single midsole cushioning unit made of EVA (Asics). (Aguinaldo and Mahar, 2003)

Shoes were also mechanically tested for stiffness values using a MTS 858 Mini-Bionix servohydraulic testing machine (Eden Prairie, MN) which was set to simulate a peak force of 2.5 times the average body weight of the participants. Ten participants (8 male, 2 female) ran at 3.23 ± 0.02m.s^{-1} across a set of three embedded force platforms recording GRF data at 1000Hz. Kinematic data was recorded by an opto-electric motion analysis system at 120Hz. The results of the mechanical testing found Shoe 3 to be the stiffest (137N.mm^{-1}), followed by Shoe 2 (124 N.mm^{-1}) and Shoe 1 (92N.mm^{-1}). The results from the GRF data reported significant differences in GRF impact peaks between Shoe 1 (1.94 BW) and Shoe 2 (1.84 BW). They also reported significant differences in GRF average loading rates between Shoe 1 (57.9 BW.s^{-1}) and shoe 2 (45.7 BW.s^{-1}). It should be noted that Shoe 3 produced the highest average loading rates but due to the variability of the data did not produce significant results to the $P < 0.05$ level. The results of these studies both demonstrate that the cushioning properties reported in mechanical tests do not demonstrate a link between cushioning and the reduction of kinetic data linked to overuse injuries, with statistical analysis suggesting increases in cushioning may expose an athlete to larger detrimental force characteristics.

A more recent study investigated the same footwear type (Asics Gel 121) with five different insoles of different shore values described by the researchers as extremely soft (ES, 35 Shore C), soft (S, 45 Shore C) , medium (M, 53 Shore C), hard (H, 61 Shore C), and extremely hard (EH, approximately 100 shore C) (Kersting and Bruggemann, 2006). Participants (n=8) ran with a natural heel strike pattern in a laboratory environment at 4m.s.$^{-1}$. GRF was measured with an embedded force platform while in-shoe force was recorded by a strain gauge based force sensor embedded into the shoe insole under the centre of the heel. The average force peaks inside the shoe however were found to be significantly larger ($P<0.05$) in the EH shoes. The next largest average impact force peak reported from the in-shoe sensor was the ES condition. An average loading rate of 58.4BW.s^{-1} for the EH condition compared to a much larger average loading rate of 72.3 BW.s^{-1} was reported for the ES condition. The authors suggest that the influence of shoe design on impact forces and other injury related characteristics are very dependent on the individual. This agrees with previously mentioned research which cited individual dependency on kinetic factors in different shod conditions (Kersting and Bruggemann, 2006). This research also provides further data that cushioning is not a simple solution for reducing detrimental impact kinetic characteristics with some more cushioned shoes reporting higher rates of loading as with the research from similar studies (Aguinaldo and Mahar, 2003). The ability of the musculoskeletal system to influence the ground reaction forces and impact shock experienced during locomotion appears to be through adjustment of movement strategy. When less cushioning is available it appears the body adjusts its movement strategy to reduce the exposure of the musculoskeletal system to forces and accelerations linked to overuse injuries. Further evidence of this is provided by research investigating participants running at 3.8m.s^{-1} on a treadmill and over ground (Hennig et al., 1996). Three different shoes all with EVA midsoles of various stiffness values were used, defined by their stiffness values by the researchers as soft (51kN/m), medium (66kN/m) and hard (341kN/m). Impact peaks reported a significantly lower peak in the hard shoe (1.8BW) compared to both the medium (2.0BW) and soft (2.1BW) shoes. Through sensory feedback the body may be adapting its movement strategy to an almost excessive point due to the perceived lack of cushioning in the harder shoes. This is demonstrated in the same study, where significantly less ($P<0.01$) of the relative load on the foot was applied to the heel area for the hard shoe (13.9%) compared to the medium (18.8%) and soft (17.1%) shoes. These results were similar to more recent research which reported that an increase in peak maximum forces in-shoe, was found to produce a reduction in the contact area of the rearfoot while increasing contact area throughout the rest of the foot (Chuckpaiwong et al., 2008). These results suggest an adjustment in the movement strategy by the participants adjusting to a more midfoot/forefoot running style. Furthermore this adds to the data suggesting that human movement strategy is affected by cushioning properties under the plantar region of the foot, leading to cushioning properties in footwear not reducing loading of the musculoskeletal system. Research by Shorten and colleagues reported that impact peaks do not give a clear indicator of cushioning effects of footwear during human locomotive movement, the study suggests that future research needs to investigate further the indicators of cushioning properties (Shorten, 2002). This agrees with conclusions drawn from this section and also suggests that any investigations into the effects of field hockey footwear should not consider simply one methodology of analysing any injury preventative factors. By reporting kinetic and kinematic effects of various footwear types further evidence can be added to the increasing amount of biomechanical studies investigating sports footwear allowing for further development and understanding of major issues effecting injury prevalence. The ideal outcome would be to find a specific design of shoe that works for all participants. However it would appear that assessment of different footwear may be required on an individual basis to biomechanically define the best footwear for an individual field hockey player.

As the data from this section mentions, an adjustment of movement strategy in different footwear conditions affects the distribution of the loading of the plantar surface of the foot. This highlights the need for footwear biomechanics researchers to investigate in-shoe plantar pressures on the foot, as landing with a more forefoot strategy may assist in reducing impact shock and GRFs but may also expose the foot to other injuries. This suggests that certain footwear construction properties and materials used may decrease the risk of one type of injury while increasing the risk of another. Considering this when an athlete is more prone to a certain type of injury it may be possible and effective in reducing injury occurrence, to select footwear that reduces the stress at a certain site that needs protecting while exposing a less vulnerable area to increased stress.

2.4.4 Footwear effects on plantar pressure

Peak localised pressure was identified previously in this thesis as exposing areas of the foot to potential overuse injuries depending on the frequency, magnitude and duration of the repetitive loading. Shod compared to barefoot conditions have been reported to reduce peak pressures over the whole of the plantar region of the foot although specific areas such as the medial arch may experience increases (Burnfield et al., 2004). Construction and supportive mechanisms in footwear as discussed in the previous section may expose the plantar region to excessive localised pressures.

Data recorded inside soccer shoes, during soccer specific movements, demonstrated characteristic pressure distribution patterns corresponding to the evaluated movements performed (Eils et al., 2004). These results suggest a greater loading of the medial and posterior parts of the feet compared to forefoot loading found during sprinting and lateral loading seen during a shot at goal. Further research found that during normal gait the most heavily loaded part of the forefoot was the first ray, with about 29% BW acting under the metatarsal head and 24% under the big toe (Jacob, 2001). These studies highlight that plantar pressure characteristics are affected by the movement strategy. The frequency of various movements within field hockey needs to be considered as although a manoeuvre may produce high localised pressure characteristics if the frequency of the movement during a match is at a low level, then it may be more effective in preventing injury or discomfort, to design shoes that would allow for the high frequency manoeuvres to be considered as the areas of importance. This would therefore suggest that for these manoeuvres, modifications that allow for a more even distribution of the loading under the foot could be beneficial. In a sport shoe it is therefore important to consider not only the explosive dynamic movements involved during the sport but also the time spent by the athlete in linear motion at varying velocities. In various running footwear types, it has been reported that peak pressures under the heel in the first 20ms after heel strike were over 5 times the sum of all the other plantar pressures (Hennig and Milani, 1995). Although this would appear high, the calcaneum is a larger bone than others in the foot such as the metatarsals and thus can withstand more stress, with most injuries caused by falls from heights (Assal and Crevoisier, 2008). This demonstrates that shoes must be designed to consider the importance of reducing pressure at specific parts of the plantar region that are known to be of particular severe injury risk when exposed to localised loading (Jacob, 2001).

Custom made insoles can be effective in the off-loading of specific localised pressure. However there is considerable intra variability between individuals. Research has reported that in customised insoles, a third reduced pressures, another third had moderate success and the final third showed no improvement (Bus et al., 2004). These findings suggest that a review after the changes have been made would be essential to be sure that the desired redistribution of pressure had been changed by the insole design. Computer simulations of the plantar region of the foot during human locomotion have been used to identify the effects of various factors on the applied pressures on the feet (Chen *et al.*, 2003;Dai *et al.*, 2006). In a computer simulation comparing a flat insole to two insoles deigned to have total plantar contact with the plantar region of the foot, it was found that peak pressure was reduced across most areas of the foot except for the midfoot regions in the total plantar contact insoles (by 19.8% to 56.8%) (Chen et al., 2003). Similar results were reported for the average pressures, with reductions in average pressure found in all areas of the foot except for the midfoot and hallux. These results would appear to be what is expected as more support is given to the midfoot which observes the largest rise in peak and average pressure, relieving other areas of the foot. This strategy would therefore be effective for reducing potentially large impact peaks in other areas of the feet experienced during field hockey play. However players with a history of midfoot injury or discomfort may be increasing their risk of injury hence prescription of such insoles should be implemented.

As well as altering footwear and insoles, previous research involving simulations found that by placing plugs made of varying shore values at the mid-foot of the in-sole of the shoe, plantar pressure could be altered (Erdemir and Piazza, 2004). The results suggested that the most effective plugs to use at the mid-foot to reduce plantar pressure without causing localised pressure at the edges of the plugs was to use larger 40mm medium-soft plugs. Furthermore the results highlighted the effectiveness of plugs that were one and a half to two times the size of the Metatarsal head. This suggested that smaller plugs may be required to reduce plantar pressure in the forefoot but they are highly material and orientation sensitive and will often require tapering to reduce the edge effects. This method may allow for minor modifications of footwear that may be suitable in terms of loading of the musculoskeletal and performance characteristics yet exposing the individual to excessive peak pressures, to be used without the risk brought about through detrimental loading of the plantar region.

The pressure distribution under the foot has been clearly identified as being influenced by the design of the insole. Similarly the design or shape of a foot therefore must be considered when measuring the interaction between the plantar surface and insole in terms of pressure.

2.4.4.1 **Footwear considerations for the effect of foot type on plantar pressures**

With various classifications of foot types (Planus (low arch), Rectus (normal) and Cavus (high arch) (Razeghi and Batt, 2002), footwear choice should consider the design of the footwear in relation to the characteristics of foot type. Research comparing normal foot type (n=30) and participants with cavus feet (n=30) found significant differences ($P<0.05$) in the peak pressure in the rear foot for the pes cavus group (Burns et al., 2005). The pes cavus group also had significantly higher ($P<0.05$) pressure–time integrals in the rear-foot, fore-foot and whole foot. In a similar, more recent study investigating plantar pressure by foot type, a group of normal foot type participants (n=34) was compared to a group of flatfoot participants (n=16) (Chuckpaiwong et al., 2008). The participants' feet were assessed by a foot and ankle orthopaedic surgeon to define if they were normal or flat feet (Ledoux and Hillstrom, 2002). A Pedar in-shoe pressure measurement system was used (50Hz, 25mm thick, 99 sensors, spatial resolution 0.391 cm^2/sensor) to measure in-shoe pressure. Participants walked (1.8 $m.s^{-1}$) and ran (3.3 $m.s^{-1}$) across a 10m runway in their own running shoes, while pressure data was recorded. Participants completed a successful trial 5 times for each speed and various kinetic data was reported. The data reported no significant differences between foot types for total foot contact area, maximum force or peak pressure. However non-significant increases were reported for flat feet in contact area and total plantar pressure. The only significant differences ($P<0.05$) reported from the plantar surface data was an increase in the contact area and peak loading of the medial midfoot in the flat foot condition which would be expected due to the low arch characteristics of the foot condition. Lateral forefoot maximum loading was significantly reduced in the flatfoot condition and peak pressure was reduced in the flat foot condition in the lateral forefoot. Although there were significant differences in the pressures experienced during human locomotion, they were relatively small and across the data, increases and decreases in plantar pressure magnitudes for each condition are seen. This would suggest that there are no obvious areas of concern when selecting field hockey footwear for flatfooted players compared to normal foot types although the results from Burns et al (2004) provide evidence that field hockey players who have a high arch (pes cavus) may be more susceptible to various overuse injuries. Knowing this, through simple foot type assessment, participants at particular risk may be identified and targeted for biomechanical assessment. This could provide information that can assist in correct footwear selection to reduce exposure to overuse injury causing kinetics for a higher risk group.

2.4.4.2 **Proprioceptive plantar feedback**

While providing cushioning to the feet during the stance phase, the insole of the sport shoe is also the interaction between the plantar region of the foot and the shoe itself. Therefore the proprioceptive feedback received from the plantar surface of the foot is directly affected by the design of the insole. Plantar sensation has been researched in many studies as a factor that may influence human movement strategy. In an extreme case where ice was used to deliberately significantly ($P<0.0001$) reduce plantar sensation, participants' movement strategy altered so that there was a significant change in the GRF experienced, with the lateral midfoot and forefoot areas exposed to increased plantar pressure (Eils et al., 2002). Although an extreme case, it demonstrates the ability of the body to gain proprioceptive feedback through the plantar region of the foot does influence kinetic and kinematic data recorded.

It has been demonstrated that altering the composition and texture of a sports shoe insole can have beneficial impacts on the resulting kinematics displayed by the participants, resulting in improved balance and thus less trips and falls and greater control over the desired movements of the participant (Maki *et al.*, 1999;Waddington and Adams, 2000;Branthwaite *et al.*, 2004;Nurse *et al.*, 2005;Waddington and Adams, 2003). Waddington and Adams (2000) found that movement discrimination scores during ankle inversion were significantly lower in barefoot compared to shod conditions for netball players in netball suitable footwear. In further research movement discrimination scores were found to be significantly worse when comparing a participant in football boots wearing socks compared to bare foot.

However by simply changing the texture of the insole to a rougher surface the movement discrimination was improved to a level similar to the barefoot conditions (Waddington and Adams, 2003). Socks with different textures made to change the sensory input at the plantar region, have also been shown to significantly change the pressure distributions under the feet (Chen et al., 1995). Another method for improving sensory feedback from the insole of the shoe is to raise the plantar-surface boundary (around the edge of the insole) using soft tubing (Maki et al., 1999). This was found to show improved stability during forward-step reactions and improved balance in many of the participants. This evidence seems to suggest that increasing proprioception in the plantar region of the foot through a rougher insole or raised boundaries may be an effective way to increase the ability of an individual to control their movements more effectively through increased neural feedback. This method may be a more effective design enhancement across a large population, compared to using raised inserts under the plantar region of the foot which may produce more variable results when comparing individuals. Within field hockey shoes it is an area that warrants further investigation particularly in cutting type movements where effective muscular control of the movement should reduce the risk of an ankle sprain. Footwear providing stability and controlling rearfoot motion may also help reduce the occurrence of ankle sprains and is discussed in the next section.

2.4.5 Footwear effects on stability and rearfoot motion

Over-pronation and increased velocity of pronation have both been linked to an increase in exercise related lower extremity injury and discomfort (Konradsen and Voigt, 2002;Ghani Zadeh Hesar *et al.*, 2009). Controlling the movement of the foot through anti-pronation footwear is suggested as a method of preventing the frequency of this injury occurring (Willems et al., 2005b). Within some sports such as field hockey, specific anti-pronatory footwear is not currently available and therefore an orthotic may be an alternative. Anti-pronatory orthotics in a similar way have been shown to reduce initial peak pronatory velocity and also the range of pronation (Nester et al., 2001).

Research investigating cushioning and balance found that when landing on softer surfaces or with thick soft soles on the footwear being used, a person will land harder, increasing the peak impact force (Robbins and Waked, 1997). According to the researchers, this was due to the participants attempting to increase their stability by compressing the soft materials in the surface and soles of the footwear to a less destabilising thinner and stiffer material. Further research has reported significantly ($P<0.05$) larger values for maximum pronation during foot to ground contact for the soft soles (-13.3 degrees) compared to both medium (-11.2 degrees) and hard (-11.1 degrees) midsoles were reported. (Clarke et al., 1983b) The shore A values of the midsoles also affected the amount of rearfoot movement with significantly ($P<0.05$) larger movements reported in the soft midsoles (17.9 degrees) compared to both the medium (16.6 degrees) and hard (15.8 degrees) midsoles. Finally the authors also reported changes in the maximum velocity between all midsole hardnesses, with soft medium and hard averages reported as 30.0ms, 36.3ms and 23.6ms respectively. The authors conclude that there may be two phases of rearfoot control and that the ideal shoe to control rear foot instability would minimise pronation and velocity of pronation. The results from Robbins and Waked (1997) suggest that increased footwear cushioning may expose athletes to higher prevalence of instability injuries in the lower extremities and may account for why increases in cushioning lead to increases in peak forces as the foot attempts to stabilise more effectively. Furthermore the evidence from Clarke and colleagues highlights more potential injury problems due to instability with increased cushioning. The results suggest an effective design of footwear for reducing injury occurrences should limit the amount of cushioning to provide increased stability.

As well as the material's thickness and stiffness in the midsole, the shape of the midsole has also been shown to influence stability. Research has reported a significant reduction in maximum pronation as the angle of the heel flare from the vertical increased from 0 degrees flare (-12.6 degrees pronation) to 30 degrees (-11.1 degrees pronation) (Clarke et al., 1983b). Total rearfoot movement also significantly reduced with increased heel flare. The results from this study investigating rear foot control appear to suggest that a shoe with a hard midsole and large heel flare would offer greater control of the rearfoot and therefore greater protection from stability injuries. It is important to also consider the influences these characteristics may have on other injury causing factors such as the loading of the skeletal system at heel strike. Only through full kinetic and kinematic analysis may shoe designs effects on potential injury causing factors be addressed.

The effects of alteration of pronatory movements associated with impact load attenuation through various midsole designs in running shoes has been previously investigated (Perry and Lafortune, 1995). Three running shoes of the same design with an EVA cushioning midsole were used. Shoe A was modified to have a 10 degree valgus wedge, shoe B was normal and shoe C had a 10 degree varus wedge.

The wedge designs were constructed with the purpose of influencing the inversion and eversion of the foot during foot to ground contact in human locomotive movement. Impact shock was measured using a shank mounted uni-axial accelerometer (1000Hz) attached to a small balsa wood plate attached to the long axis of the shank with glue and athletic tape. The participants' kinematics were recorded by a 60 Hz digital camera using a similar marker system as previous investigations (Clarke et al., 1983b). GRFs were recorded by a 1000Hz force platform embedded into the runway. Force and acceleration analogue signals were processed through a 100Hz low pass Butterworth filter. Tibial accelerations were corrected for angular motion of the tibia and gravity to allow for the proportion of the signal recorded by the accelerometer to be due to the impact shock as best possible. The study reported significant differences ($P<0.05$) between each shoe condition with the valgus shoe producing the largest rearfoot angle as would be expected followed by the normal and varus shoe. Resultant GRF impact peaks and loading rates were found to significantly increase in the varus condition. The loading rate was calculated as an average between 20% and 90% of the impact peak value. The results from the accelerometer also reported significantly larger values in the varus condition providing conclusive evidence that the varus condition would expose the athlete to increasingly detrimental GRFs. This data suggests that reduction in the ability of the foot to pronate sufficiently, which significantly increases the body's exposure to detrimental forces while increasing the maximum pronation (valgus wedge) from what would be the normal for the participant does not significantly decrease detrimental GRFs. Clearly a shoe that allows for full pronation allows a more controlled landing that allows the body to attenuate the detrimental accelerations experienced at impact with the ground. Therefore shoes with varus wedges should not be recommended for participants who may be vulnerable to overuse injuries relating to high magnitude impact shocks. This area warrants further investigation and does provide conclusive evidence for footwear used during any training involving forward locomotion only. In field hockey, where multi-directional movements are used, such a shoe may affect the stability of participants during such movement exposing them to increased risk of ankle sprains. The uppers of footwear also play a role in stability levels during such movements.

2.4.5.1 **Influence of uppers and braces on stability**

The uppers of footwear have been identified as an area for future investigation due to their possible effects on human locomotion (Morio et al., 2009). Uppers in some sports shoes such as high top basketball are designed to give extra ankle support, sports stirrup orthosis and braces are designed to perform in a similar manner. Sports shoes including any supports should allow the desired physiological range of motion but restrict excessive movement thus protecting the ankle against injury (Sharnoff, 2003). It has been reported that braced, compared to non-braced conditions, showed no significant difference in maximal eversion torque. However the study found a significant difference in the time taken to reach 66% of the maximum torque, with the braced ankle reaching this point in less time (Konradsen et al., 2005). Basketball shoes with high tops have been shown to increase stability in previous research (Petrov et al., 1988). The importance of any influence of footwear design should be evaluated by its ability to reduce injuries in a population. Ankle supports were proved to significantly decrease occurrence of injuries such as ankle sprains (Garrick and Requa, 1973). Further research has identified that athletes exposed to previous ankle sprains would be less likely to suffer a further similar injury when a sports-stirrup was used. However athletes with no history of such injuries did not demonstrate a significant reduction in likelihood of injury (Sharnoff, 2003). Burks and colleagues tested various ankle supports during research and concluded that they were all found to decrease performance (Burks et al., 1991). Further research also reported that reduced range of motion due to increased ankle support resulted in a detrimental effect on sporting performance (Robinson et al., 1986). However research has also suggested that increased proprioception enhancement in performances can be seen with ankle bracing (Papadopoulos et al., 2005). It would appear that there is conflicting evidence between studies. The results however appear to suggest that if the intervention (bracing, strapping, or uppers) does not restrict motion too much it can have a positive effect on both performance and injury prevention. For a participant particularly prone to inversion/eversion injuries of the ankle joint, high top uppers or bracing may be the best choice as reducing injury would be of a higher concern. For participants without a history of injury the evidence is not conclusive. An investigation within field hockey participants for which there is currently no published research would be useful to identify the effects of such interventions which may assist in preventing injuries and enhancing performance.

2.4.6 Force and pressure between the foot and the uppers of footwear

Injuries resulting from the uppers include in-growing toe nails and corns resulting from high localised pressure predominantly in the toe box of the shoe (Frey, 1995). Narrow toe boxes have been highlighted as being the cause of high localised pressure applied through the uppers of footwear (Rudicel, 1994). During the late stance phase of gait it has been reported that the majority of the anterior force acting on the body is not sustained at the plantar of the foot, this is also the case for the lateral forces acting in the early stance phase of gait (Hosein and Lord, 2000). The forces must therefore be acting on the body through the uppers of the shoe. A common location for corns and calluses to form is on the lateral side of the fifth metatarsal caused by interaction of the foot with the footwear (Freeman, 2002). This is an example of the affects of the shear forces acting through the uppers of footwear causing injuries through peak pressures similarly to what is reported at the plantar surface of the feet.

There is currently a paucity of literature investigating the effects of uppers on sport shoes and their effects on injury and performance regarding localised pressure as a result of shear forces. When considering the large amounts of shear forces acting on the lower extremities during sports involving dynamic multi-directional movements, the uppers of the shoes may play a vital role in comfort and injury prevention. Decreasing the pressure through footwear with extra width may help reduce the occurrence of corns in this location (Freeman, 2002). Furthermore by identifying areas of peak pressures in a similar way to the way plantar pressure is redistributed with insoles and plugs, the uppers of footwear could be designed to redistribute pressure away from problem areas. This thesis will investigate the distribution of pressure in the lateral side of the uppers in footwear (Chapter 3).

2.4.7 Footwear degeneration

Over time, cushioning properties of shoes can show a reduction. Deformation of the shoe can exaggerate any potentially dangerous movement abnormalities in the athlete, exposing them to a greater potential for overuse injuries (Hackney, 1994).

Previous work examining plantar pressure and recovery of materials during walking has been carried out by reporting a recovery time of 1300ms for a certain type of shoe (Alcántara et al., 2001). Recovery time is the time taken for the cushioning materials of the shoe to return to the same levels prior to compression. If the recovery time of a material providing cushioning within a shoe is less than the time between impacts then the shoe's performance may be detrimentally affected during sport. Through testing of the various materials used in sports footwear and referencing to the demands placed on the shoe during that sport, the use of various materials could be critically analysed. A study investigating peak pressures reported an increase of 10% after a short period of use (Eils et al., 2001). Investigating soccer shoes worn over a year of play found a lack of consistent results for the relative loads experienced in new and old shoes (Eils and Streyl, 2005). The research did highlight individual adaptation as one of the reasons behind a lack of significant changes. The effects of footwear fatigue within a single participation session may be an area that warrants investigation. Although results from some studies are not conclusive, the issue of footwear degeneration is one that has not been investigated for field hockey specific shoes. Identification of when shoes should be replaced may through kinetic and kinematic analysis factors linked to injury and performance may assist in injury prevention. Furthermore in this thesis, all footwear tested will be new at the start of the research to restrict the effects of footwear degeneration.

2.4.8 Gender specific footwear

Gender specific sports footwear including field hockey shoes, are currently commercially available. A need for distinguishing footwear between genders is reasonable as research reported that females have higher arched feet and a shorter outside foot length (Wunderlich and Cavanagh, 2001). During standing results a significant difference ($P<0.05$) in plantar contact areas was reported, when comparing genders suggesting different requirements from a sports shoe (Gravante et al., 2003). It has also been reported that footwear size selection in female populations tend towards choosing footwear that is designed for smaller feet which has been shown to be linked to injury occurrence (Frey et al., 1995). Shoe manufacturers construct slender shoes for females compared to males to allow for the more slender anatomy of the female foot. However research has found, that although the female specific shoes were reported to have provided a better fit by the female participants, the shoes did not improve the cushioning or rear-foot control characteristics during running (Hennig, 2001). Overall gender specific footwear seems sensible considering the differences reported in plantar regions and foot size. Furthermore as previously mentioned, female participants of various sports produce significantly different kinetic and kinematic data as well as being more susceptible to injury. Therefore assessing male and female participants in their gender specific field hockey footwear would be required separately to identify the specific issues for each gender that have been highlighted as being different from previous research. Within this thesis therefore, testing of the male specific footwear will be carried out with male participants only.

2.4.9 Cost of footwear

The cost of field hockey specific footwear ranges from £35-£85 (Barrington Sports, Cheshire, UK). This may have a significant effect on the choice participants make when choosing their footwear. Therefore there is a need to test whether there are any benefits from spending more on the expensive footwear in terms of influence on injury. A study of low (£40-45); medium (£60-65) and high (£70-75) cost running shoes across three brands were tested to identify any differences in plantar pressures during running (3.13 m.s-1) (Clinghan et al., 2008). The results demonstrated various differences across the ranges, but the overall conclusion of the study was that cost was not related to cushioning properties although overall the results produced lower plantar pressures in the medium and low cost footwear suggesting that if anything it was the high cost footwear that most exposed the user to injury. Further research also found there were no significant differences between high-cost, more cushioned shoes compared to the less cushioned low-cost footwear (Bishop *et al.*, 2006). The study concluded that it would need a sample of 350 participants to find a significant difference between the footwear in question. The study found significant differences in kinetics and kinematics between barefoot and shod. The research discussed in this section suggests that differences in cushioning properties may not report significant differences between typical small populations investigated in biomechanical studies and therefore make such investigations difficult to provide conclusive evidence. Within this thesis shoes of different construction and purpose will be used, the retail price of the shoes will not be a factor that will be considered when selecting the footwear.

2.4.10 Socks

As mentioned earlier, wearing socks inside footwear has been found to effect kinematic data (Waddington and Adams, 2003). A finite element model was used to simulate the effects of socks with different friction properties between the plantar region of the foot and the insole. It was found that there were significant differences in the shear forces between socks (Dai *et al.*, 2006). These findings are supported by a practical investigation in which it was found that participants (n=10) produced significantly different plantar pressure distributions whilst wearing socks with differing frictional properties (Chen et al., 1995). The results of research into the effects of socks clearly show they are a factor that needs to be considered and controlled during footwear testing. By providing participants each with the same pair of socks and ones that are similar to those that would be worn during field hockey participation the effects of socks on the data collected can be minimised.

2.4.11 Footwear considerations between field hockey participation

While hockey specific footwear and other sports specific footwear may be used during field hockey matches, skills training and strength and fitness work, there may be a need to consider movements individuals are performing in everyday life in between hockey specific movements. When the body is placed under stresses it requires suitable time to recover (Popovich et al., 2000). Sleeping provides a great deal of non weight bearing time. However field hockey players who spend a great deal of their time outside of training and matches performing weight bearing movements may be exposed to an increased risk of overuse injury. Footwear worn in weight bearing periods between participation in field hockey activities and sleep may assist in protecting from such injuries. A study was carried out investigating plantar pressures in diabetic patients (n=93) performing every day activities (Guldemond et al., 2007b). The study did not include patients if they were considered to have various conditions that would predispose them to irregular plantar pressure characteristics. The plantar areas investigated (n=6) were defined as big toe and metatarsals 1 to 5. The movements investigated were level walking, get up and go test, ramp ascending, ramp descending, stair ascending, stair descending, turning while level walking, turning while performing up and go test, turning while ramp walking, and turning while stair walking. The results produced a large amount of comparative data that reported lowest peak pressures in the 4th and 5th metatarsal regions. Walking produced the significantly ($P \leq .03$) highest peak pressure across most regions. The highest peak pressure was recorded as 275kPa under the 2nd metatarsal region of the foot with the area under the big toe (249 kPa), and 1st (248 kPa) and 3rd (264 kPa) metatarsals all reporting relatively high peak pressures during walking. General high peak pressures (>200) were reported across the regions of the foot in the big toe, and metatarsals 1 to 3, except for in turning while ramp walking and turning while stair walking. Due to the frequency of movements such as walking, footwear worn in between sports activity may assist in reducing the prevalence of overuse injury by allowing more recovery of the tissue between intense activities. This would suggest that field hockey participants that may suffer from overuse injuries in particular and spend a large part of their typical day walking, should consider footwear that may assist in relieving stresses within the lower extremities.

2.4.12 Footwear prescription

Sports shoes have a vital role to play in both performance enhancement and injury prevention. However, determining the properties of a good sports shoe appears to be a very problematic one with many factors to consider. As discussed, due to the multiple factors that affect kinematics and kinetics owing to various footwear construction and materials and the varied individual responses, the need for individual biomechanical assessment would seem apparent. To address the effectiveness of such a methodology for correct footwear selection, the rate of injury occurrence in runners who had been prescribed a shoe after biomechanical assessment (n=94) was investigated (Schwellnus and Stubbs, 2006). The biomechanical assessment group was compared to a control group of runners who had purchased shoes under normal means which may include some general advice and some individual knowledge (n=83). The runners were from similar age, weight and height groups and were the same gender. The results concluded that there was no significant difference between the two groups in terms of injury occurrence and type of injury occurring per running session. They reported that biomechanical assessment does not reduce the risk of injury. However the participants in the group who undertook the biomechanical assessment may have been more likely to have suffered a previous injury which led to them paying for a biomechanical assessment. As reported earlier, previous injury exposes an athlete to a greater chance of injury and thus this study may be an unfair test of footwear prescription through biomechanical assessment. In fact this may therefore have meant that the group with prescribed running shoes could have suffered a great number of injuries compared to the control group. Therefore prescription of running shoes may be an effective means of reducing injury occurrences. The main aim of this thesis is to attempt to investigate the effects of footwear choice on overuse injury characteristics. The outcomes may allow self prescription of footwear or furthermore demonstrate how certain testing of individuals can be used to prescribe the most effective footwear for overuse injury reduction purposes.

2.5 Conclusion

Injury risks resulting from field hockey participation may be reduced with the correct footwear. The correct footwear would be the one that reduced the detrimental loading of the musculoskeletal system linked to overuse injuries, identified through kinetic and kinematic analysis. The correct footwear must consider the movements being performed and their frequency within a specific sport and in some cases the role of the sports person within that sport. Even when all this is considered factors such as the anthropometric characteristics, age, gender, weight, and previous injuries all need to be considered. It may be that no pair of shoes will be the best for a whole population of field hockey participants. This is because research suggests that footwear factors influence kinetic and kinematics recorded in individuals differently. If this is the case, the only effective and sure way to match a sports person with a pair of sport shoes must be to include dynamic testing of the participant performing the various sporting movements involved while wearing the footwear. However by testing footwear across a population it may be possible to identify which footwear is likely to be the most effective in reducing the exposure of the participant to loading of the musculoskeletal system that may lead to overuse injuries. Furthermore, by identifying the most important factors to analyse to identify suitable footwear, a standard methodology could be developed for use in prescribing footwear to field hockey and other sports participants.

The following two chapters investigate kinetics and kinematics during multidirectional movements. Chapter 3 investigates peak pressures applied to the lateral side of the foot through the uppers of footwear. Furthermore this study aims to investigate the influence of different sports specific movements on the magnitude of the peak pressures experienced. Chapter 4 then examines the alignment of the tibia compared to the GRF vector during a cutting movement. These two chapters aim to identify areas of future research in field hockey footwear involving movements that include changes in direction. They are positioned in this thesis at the start of the biomechanical testing stage, as they do not address the main aim of the overall investigation but provide valuable information for future research comparing footwear influence.

Chapters 5 and 6 provide the basis for the kinetic testing of different types of footwear typically worn during field hockey participation. Chapter 5 investigates the relationship between GRF and tibial shock to inform the assessment of different types of footwear typically used by field hockey participants. Footwear used for further investigations is introduced in Chapter 6.

Furthermore this chapter provides information on the mechanical tests carried out on the cushioning properties of the heel area of the shoes. Results reported in this chapter, informs methodology in the following chapters.

Chapters 7 to 9 examine the influence of footwear choice on in-shoe pressure, in-shoe force and tibial acceleration during jogging and running. The results of these studies provide information on the influence of footwear on kinetics linked to overuse injuries with summary charts of all individual data provided at the end of this thesis (Tables 12.1-12.7). Chapter 10 and 11 provides a summary of the overall work and identifies further areas of research.

Chapter Three

Investigation of Localised Pressure through the Foot through the Uppers of Footwear

Aspects of this work were presented at the BASES annual conference – 2006, Wolverhampton

3 Investigation of localised pressure applied to the foot through the uppers of footwear

3.1 Introduction

Areas of intense localised pressure can lead to detrimental effects such as stress fractures of the metatarsal bones in the foot (Aerts and De Clercq, 1993) and skin abrasive conditions (Bus *et al.*, 2008;Grouios, 2004). Using a pressure measuring system (inserted inside the shoe), localised plantar pressure can be recorded over specific areas of the foot such as the metatarsal heads, mid-foot or heel during ground contact in human locomotive movement.

While studies have investigated the pressures applied to the plantar region of the foot during sport-specific movements (Bus, 2003), there is a paucity of information regarding the pressure distribution between the foot and the uppers of footwear. These non-plantar regions of the foot have also been reported to suffer from injuries linked to localised pressure (Chi and Schmitt, 2005a;Chiu and Shiang, 2007). However there are no reported values for localised pressure in these regions during normal human movement. Using a capacitive plantar pressure measuring device (PEDAR Novel, Munich, Germany), a previous study investigated the pressure between the prosthesis and the dorsal aspect of the foot in a patient who has undergone rotationplasty (Hillmann *et al.*, 2000). This surgical procedure alters the anatomical foot position, so that the foot is rotated to a vertical posterior facing position. Although this study recorded pressure distribution characteristics on the dorsal aspect, due to the altered position and function of the foot, the data recorded could not be considered for normal participants with no surgical intervention. During gait there are periods where the majority of the anterior and lateral forces acting on the foot are not applied through the plantar region (Hosein and Lord, 2000). Therefore components of these forces acting from the shoe to the foot must be applied through the upper parts of the shoe.

Previous research found that in-shoe plantar pressure data obtained during soccer specific movements showed characteristic pressure distribution patterns corresponding to the movements performed (Eils *et al.*, 2004). Studies investigating changes in directions within specific sports, such as basketball, indicate that the GRFs have reasonably large ML and AP components (McClay *et al.*, 1994). These components may result in areas of high localised pressure being exerted on the various sides of the foot by the uppers of the shoe. Research suggests that overuse injuries are a function of the magnitude of the peak pressures experienced and the frequency of these peaks (Eils *et al.*, 2004). Footwear should therefore be designed to attempt to reduce peak pressure experienced during high frequency sports specific movements.

Localised pressure formations between the foot and the uppers in footwear can result in the onset of corns and calluses which can lead to considerable discomfort and result in restricting physical activity (Grouios, 2004). Previous research has highlighted the relationship between occurrences of corns and calluses and faulty footwear, abnormal foot mechanics and high levels of activity (Freeman, 2002). This relationship becomes increasingly significant when considering diabetic sports participants, as within this population feet are a common site for medical problems (Boulton and Jude, 2004). Diabetics can suffer from both macroangiopathy (peripheral vascular disease) and neuropathy (decreased vibration and pain sensation) which expose diabetics to an increasing susceptibility to corns and calluses (Candel Gonzalez *et al.*, 2003). In a relatively small study it was found that diabetic patients suffering from ulcers, suffered 47% of ulcers on the upper dorsal of the toe/foot (Eneroth *et al.*, 2004). Decreasing the pressure through footwear with extra width or changes in the composition of the materials used in the uppers may help reduce the occurrence of injuries at this anatomical location. By devising a method to accurately record pressure distribution on any high risk areas of the foot, enhancements in protective measures may assist in reducing the occurrence of discomfort and injuries.

3.2 Methodology

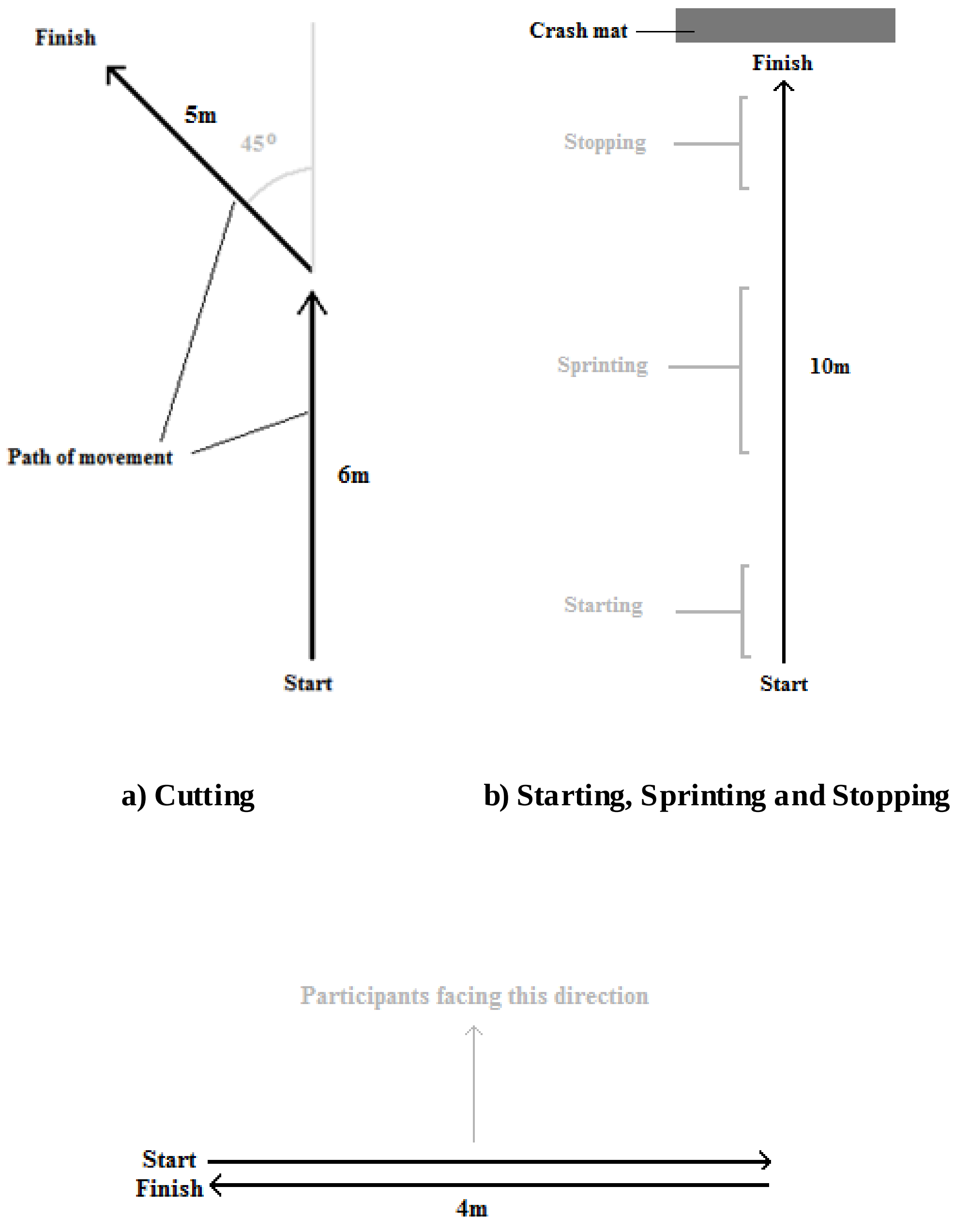
Currently there are no specific pressure measuring devices available to measure pressure across the lateral side of the uppers in footwear. In order to attempt to measure the pressure in this specific area an F-Scan 3000 (Tekscan Inc. USA) in-shoe pressure measurement sensor was used. The inserts record pressure over 4 sensors per square centimetre, and are designed for measuring in-shoe plantar pressure but can measure pressure over any area over which the sensors can be cut to shape. The insert was calibrated by using the standard method of the participant standing with the insert under the plantar region of the foot. By entering the known mass of the participant into the computer the Tekscan software is able to calibrate the system due to the change in the electrical resistance of each sensor related to the known applied force over the entire device. The insert was then cut to the desired shape (Figure 3.1) and inserted into the lateral side of the participant's sock on the right foot with small pieces of double sided adhesive tape applied to hold the sensor in place (Figure 3.2).

Figure 3.1 Adapted pressure sensor

Figure 3.2 Pressure sensor inserted into sock

To test the validity of the pressure sensor within the footwear, two healthy male participants (aged 24 and 26 years, mass 76 and 73kg respectively) with the adapted insert in place on the lateral side of the foot were used. Wearing the same pair of sports footwear (Gryphon Viper), a sphygmomanometer was wrapped around the forefoot of the shoe (dominant side). This test was conducted in a non-weight bearing condition with the participant's feet freely hanging. The pressure was exerted by the sphygmomanometer on the surface around the forefoot and was steadily increased. Force data was recorded via the sensor at set increments until the participants felt discomfort.

For the main study, six male participants (aged 26.7±2.4 years, mass 75.2±5.5 kg) with no known musculoskeletal conditions performed five specific sports movements cutting at 45º (**Figure 3.3**b), starting, stopping and sprinting (**Figure 3.3**a), and sidestepping (**Figure 3.3**c), all at a self-selected speed in a carpeted biomechanics laboratory. Each participant had the pressure sensor attached to the lateral side of their foot in their sock and wore the same footwear (Gryphon Viper). By palpating the foot through the uppers of the shoe the location of lateral border of the 5^{th} Metatarsal and Calcaneum were identified. Pressure was then applied using a pointer with a blunt head to these two anatomical landmarks (Figure 3.4). By recording the position of these points from the pressure data recorded from the sensors, it was possible to accurately reference the lateral side of the participant's right foot in contact with the pressure sensor (Murphy, 2006, personal communication). This facilitated the recording of peak pressures within the areas under investigation to be consistently recorded across the trials. Ten trials of data were recorded for each movement which according to previous research would provide a suitable statistical power for the amount of participants recruited (Bates *et al.*, 1992).



a) Cutting **b) Starting, Sprinting and Stopping**

c) Sidestepping

Figure 3.3 Participants movement strategies for (a) cutting, (b) starting, sprinting and stopping and (c) sidestepping

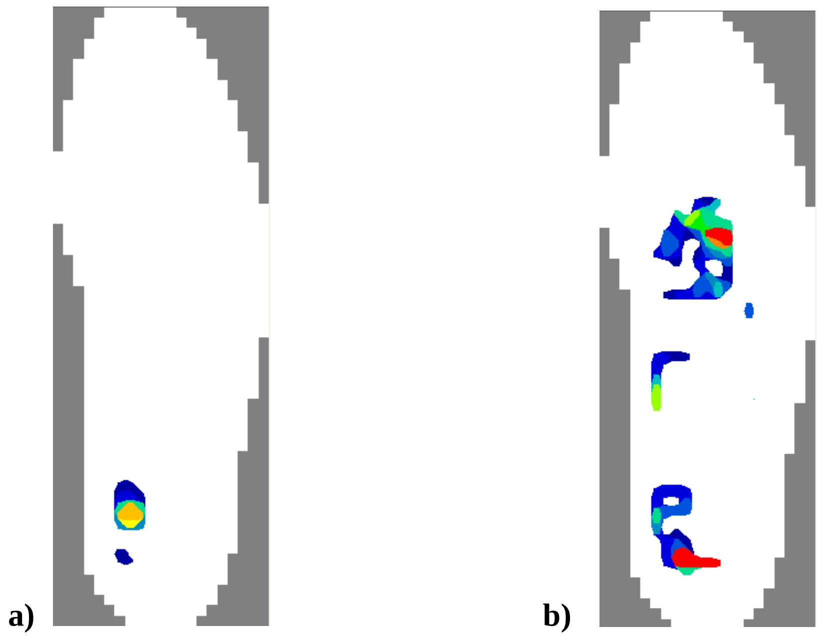
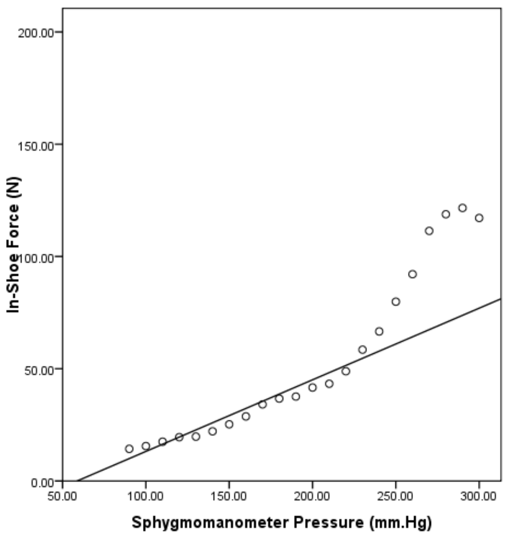


Figure 3.4 Typical pressure distribution patterns: a) applying pressure with a blunt pointer to the lateral side of the calcaneum b) during a cutting movement.

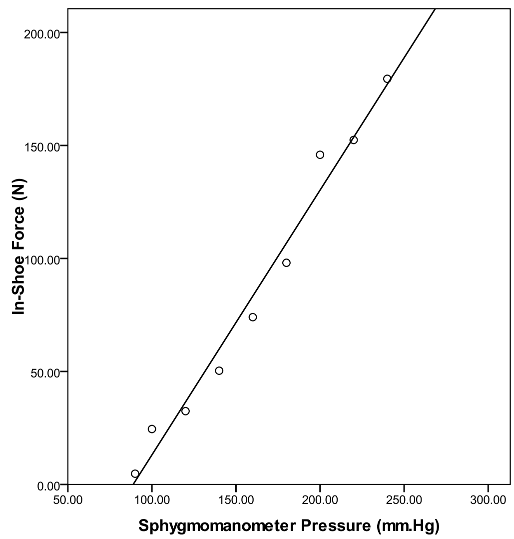
3.3 Results and discussion

As shown in Figure 3.5, the results from the sphygmomanometer test indicated a steady increase in the force recorded by the sensor for each increase in the pressure applied by the sphygmomanometer over the forefoot. Bivariate regression analyses were performed to compare the relationship between the force measure by the in-shoe sensor and the applied pressure from the sphygmomanometer. As one variable was known and one measured r^2 values were calculated using a Pearson's correlation with significant values ($P<0.001$) of $r^2=0.98$ (Figure 3.5a) and $r^2=0.89$ (Figure 3.5b), reported for participant 1 and 2 respectively. This test demonstrates the ability of the adapted in-shoe pressure to measure increases in pressure over specific areas of the lateral side of the foot and the uppers of the footwear.

Although the increase in the applied force was larger depending on the participant, this can be accounted for by the different shapes of the two participants' feet resulting in a larger proportion of the applied pressure from the sphygmomanometer being applied to the sensor. This small study does suggest that much like plantar pressure, when the same amount of force is applied over the same sized area then the distribution of this force can differ between participants.



a) Participant 1



b) Participant 2

Figure 3.5 Force recorded by the in-shoe pressure sensor under applied pressure from an externally attached sphygmomanometer for two participants.

The results of the main study recorded pressure distribution patterns which are dependent on the movement being performed (

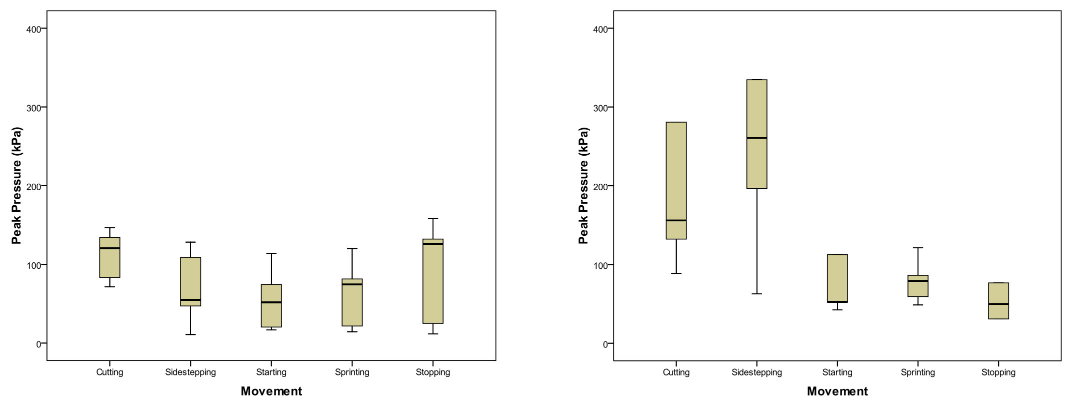
Figure 3.6). Higher peak pressures are clearly present in the movements involving a specific lateral component (cutting and side-stepping). It was clear from observations of the pressure distribution patterns recorded that during the lateral movements (cutting and side-stepping) areas of high peak pressures were prevalent at the lateral side of the 5th metatarsal head and the lateral side of the calcaneum. Mean and standard deviations were calculated over the ten trials per participant for each movement and are presented in Table 3.1. As the magnitude of localised pressure is the important factor in assessing the possible injury risk, the results were not normalised to bodyweights of the participants to allow clearer identification of the values reported.

Table 3.1 Mean in-shoe peak pressure recorded during sports specific movements at the lateral side of the foot for all participants.

	Calcaneum (kPa)		5th Metatarsal (kPa)	
Cutting	112.8	±29.2	182.4	±81.1
Sidestepping	67.5	±43.5	241.6	±109.5
Starting	54.9	±39.0	74.5	±35.1
Sprinting	64.4	±39.9	79.0	±25.1
Stopping	96.6	±62.0	66.4	±49.0

This data in Table 3.1 demonstrates similar peak pressures reported in the plantar region of the feet during locomotion (Burnfield *et al.*, 2004). This finding agrees with previous research that during human locomotion, a large proportion of the reaction forces are acting on the skeletal system through the uppers of the footwear (Hosein and Lord, 2000).

An ANOVA test was used to compare the means. Significant ($P<0.05$), differences were found between the peak pressures reported at the lateral side of the calcaneum and the 5th metatarsal head. Larger mean peak pressures were found to be applied to the lateral side of the 5th metatarsal head in most of the conditions with only the stopping movement reporting a larger mean peak pressure at the side of the calcaneum. Between the movements an ANOVA test reported no significant differences ($P>0.05$) between the mean peak pressures applied to the lateral side of the calcaneum when comparing the effect of the movement. There were however significant differences reported ($P<0.05$) at the lateral side of the 5th metatarsal head with the sidestepping movement exposing this area of the foot to significantly larger peak pressures than those measured during starting, sprinting and stopping (Figure 3.6). Furthermore, significance value of $p=0.074$ was reported when comparing the same variable between the cutting movement and the stopping movement.



a) Lateral side of the calcaneum

b) Lateral side of the 5th metatarsal head

Figure 3.6 Mean in-shoe peak pressure recorded on the lateral side of the foot for all participants.

These results show that the pressure in the uppers can be relatively higher in different areas between the foot and the uppers of the shoe. Most notably during sidestepping where the mean peak pressure recorded during sidestepping (241.6 kPa) was 360% larger than the peak pressure between the upper of the shoe and the side of the calcaneum (67.5 kPa). This would suggest that during sidestepping, most of the traction between the shoe and the surface would be occurring in the forefoot plantar region.

For sports involving a high frequency of the movements investigated over a sustained period of time there is a need to consider the distribution of the pressure in the uppers of the footwear used. Reducing this pressure on the feet may help reduce discomfort, skin abrasion and the onset of corns and calluses (Freeman, 2002).

Adaptations to insoles have been identified as being effective in reducing specific localised peak plantar pressure values (Erdemir and Piazza, 2004). Similar adaptations to the support provided by the uppers of field hockey footwear may have similar effects on the pressure distribution applied to the foot through the uppers. Previous research has used real-time plantar pressure feedback to enable the participant to adapt their movement characteristics during normal gait (Femery *et al.*, 2004). A similar system may help field hockey players performing movements placing them at risk of high lateral pressure on the side of the foot to adapt their movement strategies to prevent exposure to high lateral pressures. Field hockey players will only spend short times of the game performing multiple sideways movements. However during training such movements may be more frequent.

Furthermore as pressures are found to be high on the lateral sides of the feet in footwear during sideways movements it would suggest that this would be the same for the anterior edges of the metatarsals during stopping with potentially larger peak pressures being experienced. This is an area that warrants further investigation although during this investigation it was not found to be possible with the technology available to place pressure sensors inside the anterior end of the shoe to record such data due to the shape of the shoes and the sensors setup.

3.4 Conclusion

This study has recorded localised pressure between the lateral side of the foot and the uppers of a shoe. This area is reasonably accessible using an adapted in-shoe plantar pressure device. Other regions of the foot such as the front of the toe-box are more difficult to access and record reliable data from, due to the shape of the region involved and the devices currently available. The results from this study show the foot experiences high levels of localised pressure between the uppers and the foot during field hockey specific movements. Through development and implementation of this technique, improvements in footwear injury prevention and comfort characteristics may be possible within field hockey specific and other types of sports specific footwear. By designing shoes with supports on the lateral sides of footwear in places where the foot experiences lower levels of peak pressures such as possibly the lateral side of the midfoot, peak pressures in the uppers may be decreased by redistributing a larger component of the force away from the areas at most risk. This design feature may be particularly useful for any field hockey players who have suffered previous injuries in a specific area of the uppers. By choosing the correct shoe design or maybe placing gel pads within the uppers of the shoes it may be possible to alleviate the peak pressures experienced in the area of concern allowing the athlete to reduce their injury risk while reducing the detrimental effect the discomfort may have had on their movement and thus their performance.

Movements involving sideways components such as the cutting movement investigated in this chapter are more complex than normal forward locomotion, with larger components of the GRF applied horizontally. While this Chapter has investigated the effects of such forces through the uppers of the footwear, investigating the loading of the lower extremities with the resultant GRF components may highlight injury potential and how footwear characteristics can affect this. The following chapter aims to investigate this area of research.

Chapter Four

Influence of Insole Texture on Tibial Alignment with the Ground Reaction Force

This section has been published in Clinical Anatomy, 2006

Aspects of this work were presented at the BACA scientific meeting - 2006, Keele

4 Influence of insole texture on tibial alignment with the Ground Reaction Force

4.1 Introduction

Impact forces occurring during the first 50ms of the foot to ground contact phase result in loading of the tibia which has been linked with various overuse injuries (Lake, 2000;Nigg, 2001;De Wit *et al.*, 2000;Mizrahi *et al.*, 2000a), However, increases in the frequency and magnitude of loading on the human skeletal system have been found to increase the bone mass and strength at the site of loading (Egan *et al.*, 2006;Cullen *et al.*, 2001). This suggests that impact loading has a positive effect on the development and maintenance of bone although excessive and prolonged loading can result in stress fractures (Nigg, 2001). A stress fracture of a bone occurs when a bone is placed under repeated loading and unloading forces. Stress fractures often develop over a long period of time within the tibia, in which cases the athlete may experience some discomfort and localised pain (Ekenman *et al.*, 2001). Risk factors for such injuries in the lower extremity have been cited as intensity and duration of sporting activity (Micheli, 1986). The higher the frequency of the loading and unloading of the bone, the lower the value of stress applied to the bone has to be in order to result in injury (Nigg *et al.*, 2000).

The GRF is recorded as three components Fx, Fy, and Fz (ML, AP and Vertical), calculating the sum of these three components allows the study of the resultant GRF. The resultant GRF therefore has a single magnitude and direction so can be described by a single vector. Although the GRF is distributed over the sole of the foot in contact with the ground, a point known as the centre of pressure (CoP) can be considered the point of application of a resultant GRF vector (Cavanagh and Lafortune, 1980). This is the point at which the sum of all the vectors acting from the force plate intersects the sole of the foot or shoe (Nigg and Herzog, 1999). The CoP can therefore be used when investigating all the components of the GRFs.

In a sporting situation which involves a forward running motion, the vertical component of the GRF is much larger than the ML and AP components leading to studies only considering the vertical GRF component (De Wit *et al.*, 2000). Researchers found that during a cutting movement the mean peak ML, AP and vertical components of the resultant GRF vector were 1 BW, 1.1BW, and 2.3 BW respectively, while values of 0.2 BW, 0.4 BW, 2.5BW, for the ML, AP and vertical components were reported from the same athletes during running forwards in a straight line (McClay *et al.*, 1994). These results mean that during a cutting movement the orientation of the resultant GRF will be more horizontal than during running forwards.

Multidirectional movements in hockey such as cutting may therefore pose a potential injury to the lateral sides of the feet as identified in the previous chapter. Furthermore, they may also increase the risk of an overuse injury in the tibia compared to forward running. This is due to a movement strategy that may expose the participant to potentially damaging loads at a relatively large angle to the longitudinal axis of the tibia. The alignment of the lower extremities has been identified as an area of concern for overuse injuries (Wen et al., 1997). Early research suggested that, a limb in normal alignment may effectively withstand the repeated loading apparent during running (Cavanagh and Lafortune, 1980). When a bone experiences a force not acting along the axis of the bone (known as a bending force), the total stress on the surface of the bone can be multiple times larger than the same force acting along the long axis of the bone (Nigg and Herzog, 1999). Clinical and experimental investigations provide evidence that stress injuries occur at the site at which the maximum tensile stress due to bending occurs (Mizrahi *et al.*, 2000b;Daffner, 1984). The bone's ability to resist bending moments has been identified as a factor that can reduce the risk of a stress fracture occurring (Milgrom et al., 1989). This ability of an individual's bones may be a cause of variability between individuals to avoid such injuries. It is clear that the alignment of the various bones in the human body compared to the direction of the applied force will influence the onset of injury. The body's ability to orientate its lower limbs to align the applied resultant force along the long axis of the tibia could reduce the likelihood of developing an overuse injury. Research has identified the alignment of the skeleton as one of the most important factors to consider when designing running shoes and Orthotic inserts (Nigg, 2001). Insole texture has been reported to influence the proprioception of the musculoskeletal system through the plantar region of the foot (Waddington and Adams, 2003). Socks with different textures made to change the sensory input at the plantar region, have also been shown to significantly change the pressure distributions under the feet (Chen et al., 1995). A further aim of this thesis is therefore to identify how the alignment of the tibia and the resultant GRF vector during a hockey specific cutting movement are influenced by plantar sensory input conditions. By calculating the applied angle of the resultant GRF to the long axis of the tibia, various loading characteristics with links to overuse injuries can be investigated in shoes with different insole surfaces. This is a newly developed method of analysing kinetic data compared to kinematic data and may provide the basis for further research using this developed methodology.

During a cutting movement the resultant GRF vector is more horizontally orientated due to the larger ML and AP components. Considering the GRF to be acting on the tibia and fibula through the ankle joint, a reasonably vertically orientated shank would expose the tibia to a bending force. This would be applied by the relatively large component of the resultant GRF vector that would be perpendicular to the shank. By selecting properly cushioned footwear, the risk of acquiring injuries associated with repetitive impacts may be reduced (Lake, 2000). Various adaptations such as insole surface texture and tubing around the borders of the insole can be made to assist in the foot’s ability to provide neural feedback (Waddington and Adams, 2003). Increased neural feedback will give the body more information allowing for improved movement strategies. These adaptations can be very dependent on the individual athlete using the shoes (Gillespie and Dickey, 2003).

4.2 Methodology

A single participant was used for this study as the aim of the study was to introduce a new method of analysis and identify if the in-shoe texture could influence the movement in an individual. The participant was a healthy male aged 23 and mass 70kg was used for this study. Reflective markers were attached at specific anatomical points on the participant to allow for the plug in gait full body model to be created from the kinematics recorded (Figure 4.1).

The three components of the resultant ground reaction force vector were measured using a strain gauge force platform (AMTI Inc, USA). Kinematic data was sampled at 100Hz using an 8 camera opto-electronic motion analysis system (Vicon Peak, UK). The participant was required to perform a cutting movement with their right turning foot contacting the force plate during the cutting movements. Markers were placed either side of the force plate to allow the participant to practice the cutting movements with approximately a 45 degree turn (Figure 4.2).

The participant had a pair of smooth insoles inserted into their shoes and was then given time to warm up and practice the cutting movements until they felt comfortable and could repeat the skill successfully filling the criteria of landing on the plate with their right foot and cutting at the desired angle. The participant performed a number of cutting movements at self-selected speeds while kinetics and kinematics were recorded at 1000Hz and 100Hz respectively, until eight good sets of data were recorded. The smooth in-sock (top cover of an insole) was then replaced with a rough in-sock and once the participant was comfortable they performed the cutting movements a number of times once more until ten sets of good data were recorded. The kinematic data was smoothed using a Woltring filter and the angular difference between the tibia and the resultant GRF vector can be seen in Figure 4.3 where the red arrow represents the resultant GRF vector with its origin at the CoP. The angular difference was calculated for each frame of data. A one-way ANOVA was used to test at the significance level $P<0.05$, the difference between various GRF and angular characteristics.

Figure 4.1 Marker placements on right side lower limbs

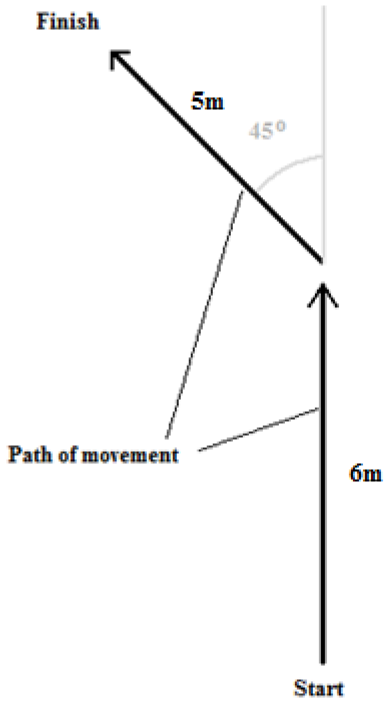


Figure 4.2 Path of cutting movement performed by participant

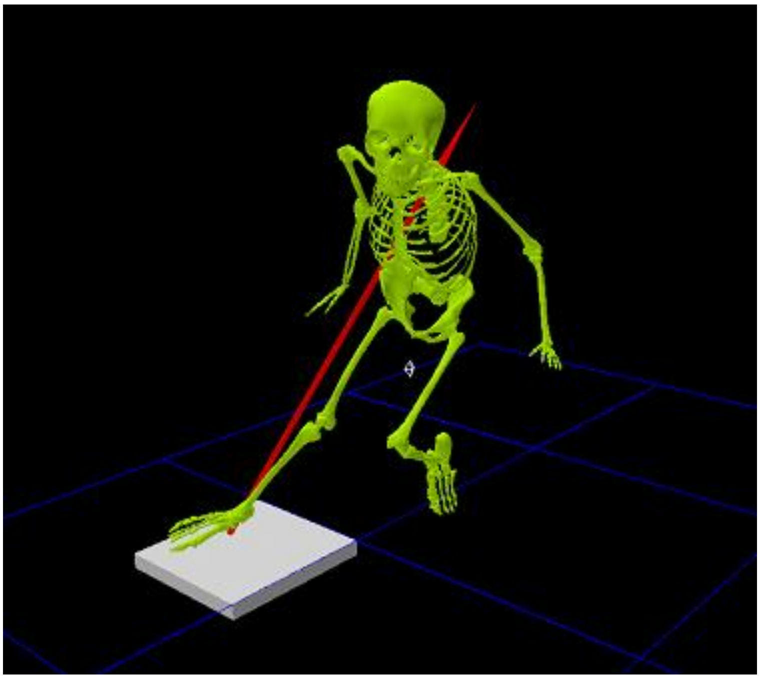


Figure 4.3 Typical resultant GRF vector compared to skeletal position during a cutting movement

4.3 Results and discussion

The GRF components recorded typically showed similar characteristics as found in the study by McClay *et al* (1994), with the ML component typically being about half that of the vertical component (Figure 4.4). This meant that the direction of the applied resultant GRF was at an angle deviated from a typically vertically orientated GRF vector seen during forward running data.

Comparing the effects of the rough and smooth insoles on the force and loading peaks and the angles between the tibia and the GRF at these points provided mostly inconclusive data (Table 4.1). However there was a significant mean difference of 8.88º ($P<0.05$) recorded when comparing the angle between the tibia and resultant GRF at the 1st impact force peak. This suggests that for this participant, the change in the insole texture did affect the kinematics recorded. The change in the angle at peak force would appear to reduce the bending force applied to the tibia and thus may be a beneficial effect in protecting the lower extremities from injury for the individual.

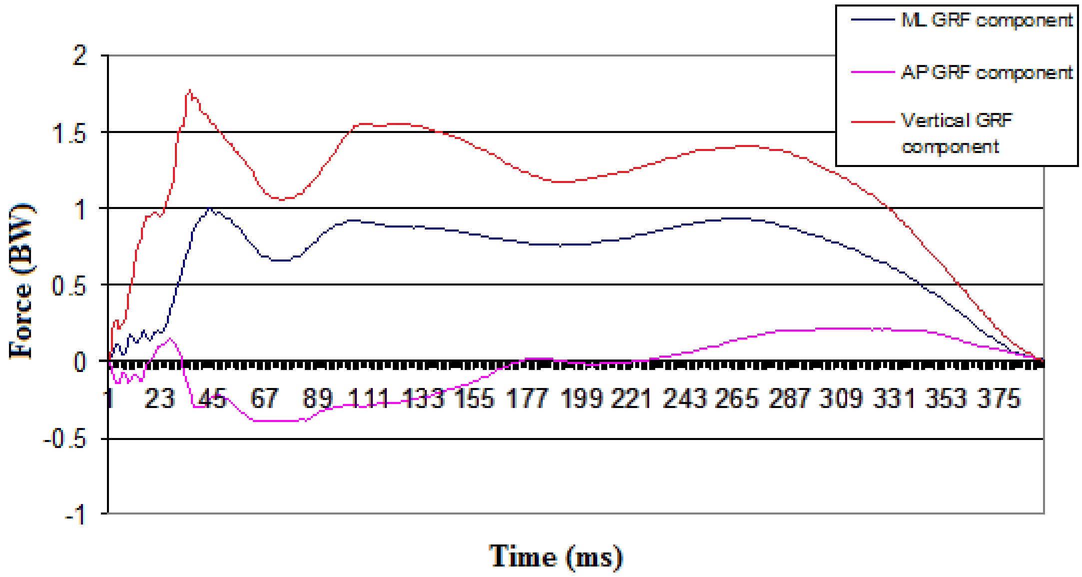


Figure 4.4 Typical GRF components during a single trial using smooth insoles

The average angular difference between the tibia and the applied GRF from both conditions was 30.35º, which suggests a large component of the GRF is not being applied along the axis of the bone leading to an exposure to potential injury risks (Daffner, 1984;Mizrahi *et al.*, 2000a;Milgrom *et al.*, 1989;Nigg and Herzog, 1999;Mizrahi *et al.*, 2000c). The technique used in this study to identify and consider the angular difference between the tibia and the resulting applied GRF is being suggested here as a methodology for identifying possible injury causing movement characteristics.

Table 4.1 Mean results of kinetic and kinematic data during cutting movement

Variable	Smooth	Rough
Time to Peak GRF (ms)	43.8 ± 5.0	42.1 ± 4.4
Max GRF (N)	1309.8 ± 110.9	1359.6 ± 80.2
Angle between Tibia and Resultant GRF at Impact Force Peak (degrees)	34.05* ± 6.6	25.17* ±2 .9
Time to Peak Loading Rate (ms)	24.8 ± 17.1	15.7 ± 12.7
Peak Loading Rate (KNs^{-1})	88.2 ± 26.7	123.1 ± 19.3
Angle at Maximum Loading Rate (degrees)	33.29 ± 9.5	34.11 ± 4.3
Mean Difference of the Angle Between tibia and Resultant GRF and Tibia (degrees)	33.3 ± 8.3	27.4 ± 1.2

*=significant difference ($p<0.05$) between rough and smooth condition

When considering the findings of this research to sporting application, the frequency of a movement pattern within participation needs to be identified when investigating the potential for the occurrence of an overuse injury. If load peak magnitudes are high enough and at a sufficient frequency, the bone will become unable to self repair resulting in injury (Nordin *et al.*, 2001). Some sports such as basketball and tennis require the athletes to perform side stepping movements at a high frequency over long periods of time. Field hockey involves shorter periods of play where such movements are performed. However during training repeated dynamic lateral component movements may be observed and be a potential injury risk factor. The results from this study suggest that the magnitude frequency and direction of GRFs for such movements require consideration if overuse injury prevalence is to be reduced.

One of the major limitations of this study is the movements' recorded dissimilarity to those occurring during field hockey participation. Performing a cutting movement in laboratory conditions may differ greatly from a similar movement produced during a sports situation For instances when the sportsperson has to make the decision to move in the desired direction as a result of an external factor, such as a change of direction by an opponent. The laboratory situation allows for the participant to prepare and have prior knowledge of the movement they are to perform. The effect of force plate targeting which has been shown to influence movement characteristics (Challis, 2001) was kept to a minimum as the participants practiced and paced out the movement until they could successfully strike the plate without any noticeable targeting of it. Trials where the participant did look down at the plate or appeared to reach for the plate were discarded. Replication of more randomised movements may produce more real life movement characteristics. However recording the kinetic data may prove difficult as the foot to ground interaction requires the entire foot to be in contact with the force platform used. There is a need for further research to assess the validity of the techniques used within this study. Investigating other typical human movements such as sprinting, stopping and sidestepping may be valuable for identifying possible overuse injury potential.

4.4 Conclusion

The insole texture does appear to have an effect on the movement strategy of the participant used in this study. The effect of the insole suggests that a rougher insole surface may help place the lower extremities under lower bending forces and thus decrease the risk of overuse injury. As this was a single participant evaluation, a similar study with a larger cohort of participants in line with guidelines produced by Bates *et al* (1992) will give more conclusive information on the effects of insole texture and the kinetic and kinematic data investigated in this study. This methodology for investigating footwear effects on the alignment of the musculoskeletal system and the resultant GRF vector needs to be investigated further. In the future, by comparing kinematic and kinetic data (as in this study) of athletes with histories of overuse injuries at the tibia, it may be possible to identify if the alignment of the resultant GRF vector and the tibia are a factor in the prevalence of overuse injuries.

Chapters 3 and 4 have investigated multi-directional movements. However, the majority of field hockey participation will involve movement in a predominantly forward motion. The repetitive nature of such movements exposes the musculoskeletal system to overuse injuries as discussed in Chapters 1 and 2, in particular stress fractures at the tibia and other areas of the lower extremities. The next chapter investigates the relationship between GRFs and tibial acceleration to identify characteristics that may be linked to overuse injuries.

Chapter Five

Impact Forces Relating to Tibia Impact Accelerations

5 Impact forces relating to tibia accelerations

5.1 Introduction

There have been many studies investigating detrimental characteristics of the human body's impact with the ground during human locomotion. When not excessive in terms of frequency and magnitude, loading the musculoskeletal system provides essential health benefits, including maintaining a suitable level of bone density (Vico *et al.*, 2000;Bennell *et al.*, 1996a). However when the magnitude or frequency of impacts are excessive, overuse injuries such as stress fractures have been associated with the transient impact shock (Milner *et al.*, 2006).

Measuring the impact shockwave transmission through the skeletal system at the tibia has been carried out successfully using rigid attachments invasively attached directly to the bone itself (Hennig and Lafortune, 1991;Lafortune and Hennig, 1991;Lafortune *et al.*, 1995a). However, this methodology causes much discomfort and requires invasive surgical procedures, thus is not practical in many situations. The data from such studies has been compared to skin mounted accelerometers which provide a non-invasive method of estimating the actual tibial shock values. Large differences have been found between the signals for skin and bone mounted accelerometers. However, it has been shown that through the use of a low pass filter, the large component of the signal present due to the skin interaction between the bone and accelerometer, can be reduced to a level where a good estimation of the bone acceleration can be recorded (Shorten and Winslow, 1992). The skin mounted accelerometer methodology has been used in previous research, with the accelerometer attached tightly to the skin at the anterior medial aspect of the tibia (Coventry *et al.*, 2006;Flynn *et al.*, 2004;Pohl *et al.*, 2008;Laughton *et al.*, 2003;Milner *et al.*, 2006). This position provides minimal skin interaction and minimal effects of acceleration due to the angular motion of the tibia about the ankle joint.

During human locomotion in shod conditions, research has found there is a decrease in the first impact peak when more cushioning is present through the soles of the footwear (Hennig *et al.*, 1996). However the time to reach the impact peak was reported to be much shorter producing a higher rate of loading which has been strongly correlated to increases in impact shock measurements (Hennig and Lafortune, 1991). Throughout the studies investigating the impact phase of human locomotive movement there are many variables that have been identified as relating to injury such as the initial impact peak, average loading rate (measured in various ways), instantaneous loading rate and time to peak loading rates (Hennig and Lafortune, 1991;Nigg *et al.*, 1988;Perry and Lafortune, 1993;Guido *et al.*, 2009;Laughton *et al.*, 2003;Diop *et al.*, 2005;Kong *et al.*, 2009;Pohl *et al.*, 2008). Various ways of calculating the rate of loading have also been used in previous research. One methodology calculated the average loading rate from 20-80% of time to impact peak (Laughton *et al.*, 2003). Similar methods calculating from 20-90% have also been used (Perry and Lafortune, 1995). A method used by Munro and colleagues calculated the loading rate from 50N, to BW plus 50N (Munro *et al.*, 1987). Calculating a loading rate this way does not require identification of a force peak and may therefore be a simpler characteristic to use and prone to less human error. The same is also true for studies that reported the instantaneous loading rate (Bus, 2003;Laughton *et al.*, 2003;Bergmann *et al.*, 1995). By calculating the maximum difference found between each sample of GRF data, a peak loading rate during the impact phase can be found and recorded. The timing of the peak loading rate has not been examined by many studies and may be a factor that provides further information on the injury potential of GRFs. From the methodologies used in previous studies it is not clear if there is a conclusive way of analysing force data as a predictor of impact shock. A comparison of all the methodologies may provide evidence to allow identification of the best methodology to use.

Accelerations recorded from shank mounted accelerometers used to directly measure impact shock are an effective method to compare footwear. As the sensor can be left attached and in a secured position between trials in different footwear, differences due to the attachment and position of the accelerometer can be minimised allowing for an effective comparison between the shoes for each participant.

Material test results have previously found significant differences between footwear but when the same footwear was tested during human locomotion measured with shank mounted accelerometers, no significant differences were reported (McNair and Marshall, 1994). This highlights the complexities of human locomotion on the accelerations experienced in the lower extremities and further knowledge of the factors affecting the detrimental forces relating to these accelerations warrants investigation.

5.2 Methodology

Thirteen adults (Age 30.0 $\pm$ 9.4 years; Height 1.74 $\pm$ 0.06 m; Mass 70.6 $\pm$ 8.1 kg) comprising of 10 male and 3 females, volunteered to take part in this study. All were injury free at the time of data collection and completed an informed consent form. Participants were required to run between two sets of timing gates positioned 4m apart and either side of the force platform. The participants had a 10m run up to the plate with 10m after the plate to slow down. They were instructed to run through the second set of gates before slowing down (Figure 5.2). A thick crash mat was used against the end wall to allow the participants to stop safely and reduce the risk of injury through collision with the wall. Each participant was required to perform 8 good trials. A trial was considered good when the participant landed with their right foot fully in contact with the force plate with no observable adjustments made to target the force plate. Participants were required to run at $4 m.s^{-1} \pm 5\%$ measured by the timing gates.

A tri-axial accelerometer (Biometrics ACL300) was mounted to a lightweight carbon-fibre plate via a securely glued lightweight bolt and thread attachment. The total weight of the accelerometer and mounting system was 13g. The carbon-fibre plate was firmly attached to the shank via surgical adhesive tape. By using skin stretching techniques the plate was attached tightly so the accelerometer was positioned on the distal anterior-medial aspect of the tibia and 8cm above the medial-malleolus. The accelerometer was orientated to measure the acceleration along the longitudinal axis of the tibia (Figure 5.1). This accelerometer and attachment system was the same used in a recent publication investigating differences in fencing footwear (Sinclair *et al.*, 2010). By positioning the accelerometer near the malleolus the radius of the motion of the sensor about the ankle joint was minimised.

Figure 5.1 Accelerometer attached to the antero-medial aspect of the shank

1000g measurement) signal was set to 100mV/g providing a measurement range of ±100g. The sampling frequency was set to 1000Hz. The Analogue Data signal was recorded through Qualisys Track Manager software (OMG, Oxford), via a biometrics data collection device attached via a 20m wire. Force data was recorded through a force plate sampling at 1000Hz embedded in the ground of the biomechanics laboratory. The analogue signal was recorded simultaneously with the accelerometer data through Qualisys Track Manager. The accelerometer signal was processed through a Butterworth low-pass filter set to 60Hz. This filter was used to exclude the component of the signal due to skin artefact and the resonance of the device,

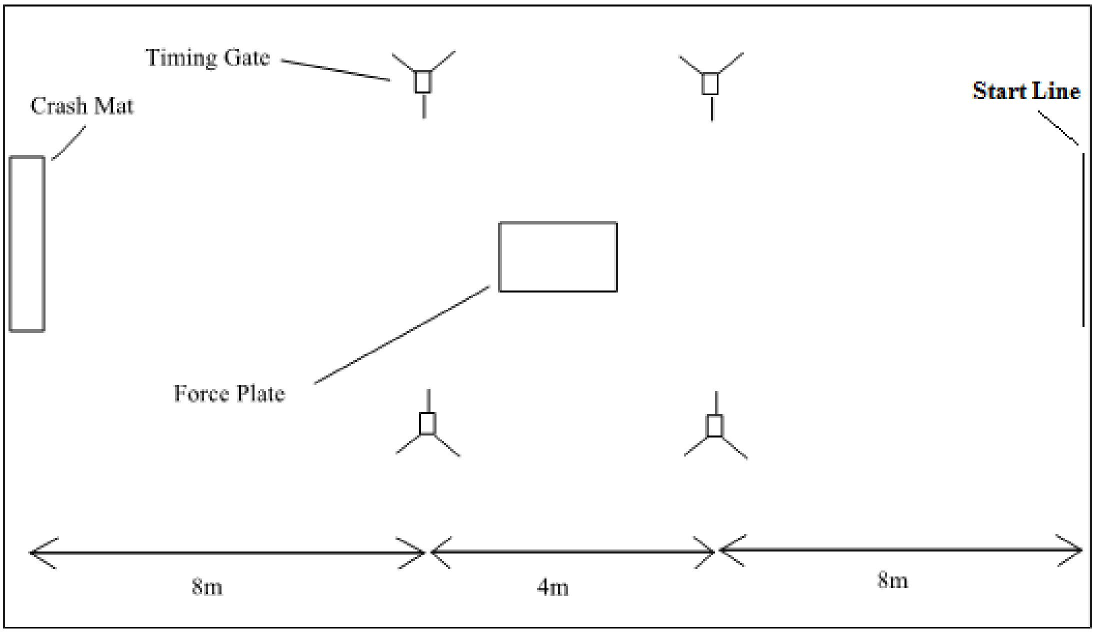


Figure 5.2 Setup of the biomechanics lab for data collection

in line with the findings from previous research (Shorten and Winslow, 1992).

Multiple bivariate regression analyses were performed to compare the relationship between the various GRF characteristics and the tibial acceleration measured. Reporting the co-efficient r value as both variables were measured to allow comparison with previous research (Laughton *et al.*, 2003;Hennig *et al.*, 1993).

5.3 Results and discussion

The VGRF (Figure 5.3), vertical loading rate (Figure 5.4) and tibial acceleration (Figure 5.5) data generated peaks with characteristics as would be expected, including peak tibial accelerations and loading rates occurring prior to the identified first vertical force peak. A first impact peak was easily identifiable for most of the data recorded, however in some cases where there were double peaks or only some minor deformation of the vertical-force time graph it was necessary to take a best estimation of the impact peak from the graphed data. The mean impact force peak values reported in Table 5.1 were similar to those from previous human locomotion research (Cavanagh and Lafortune, 1980;McClay *et al.*, 1994;Kersting and Bruggemann, 2006).

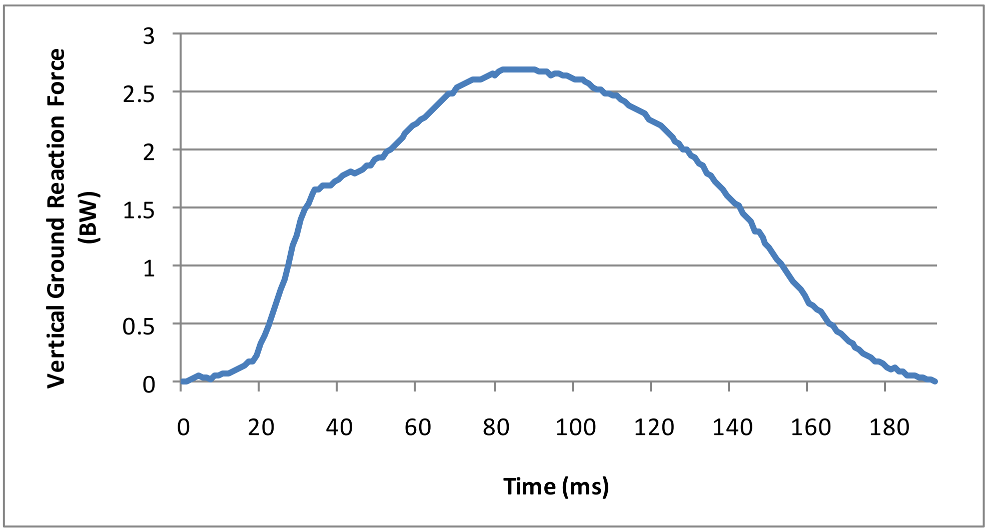


Figure 5.3 Typical vertical force data during stance

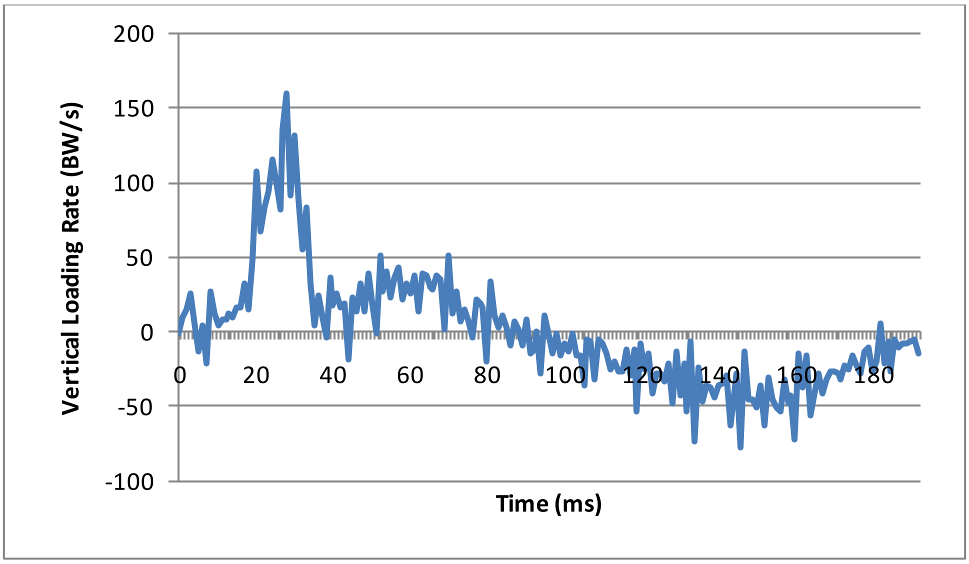


Figure 5.4 Typical vertical loading rate data during stance

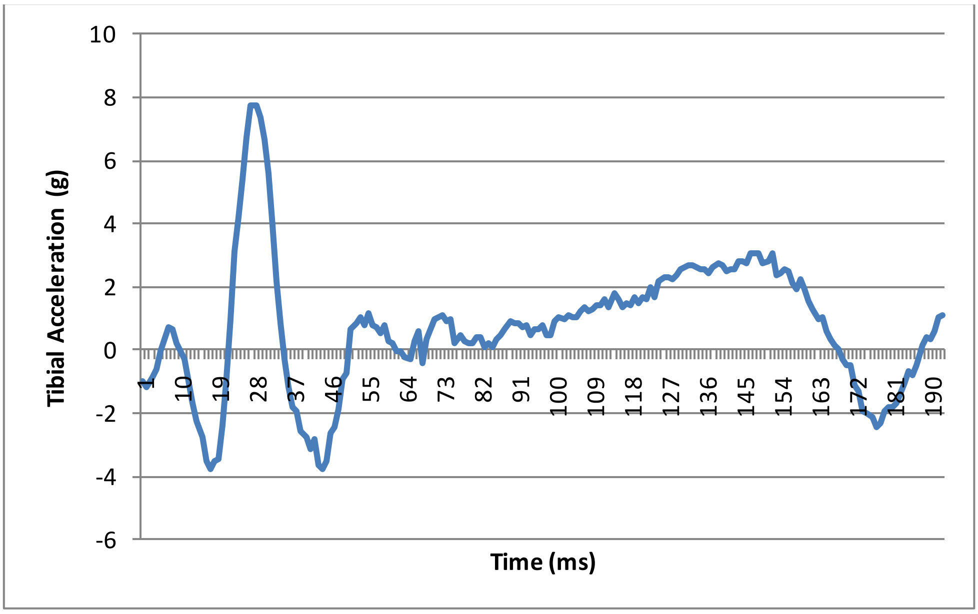


Figure 5.5 Typical tibial acceleration data during stance

Table 5.1 Mean GRF and tibial acceleration values from all participant's data

Variable	Mean Value
1st Vertical Force Peak (BW)	2.56 ±0.3
2nd Vertical Force Peak (BW)	2.77 ±0.38
Average Loading Rate (BW.s-1)	106.7 ±26.4
Peak Vertical Loading Rate (BW.s-1)	246.9 ±61.6
Peak Tibial Acceleration (g)	9.5 ±3.3
Time to 1st Vertical Force Peak (ms)	25.3 ±5.4
Time to Peak Vertical Loading Rate (ms)	15.9 ±4.3
Time to Peak tibial acceleration (ms)	21.7 ±5.9

The data reported across the participants for the timings of events (Table 6.2) shows a slightly stronger negative correlation for the time to peak loading rate ($r=-0.36$, $P<0.05$) than the 1st impact peak ($r=-0.34$, $P<0.05$). However, in the individual participants, three reported strong significant correlations for the time to 1st force peak compared to just one in the time to peak loading rate.

The stance time has a very weak correlation across the population however in five of the participants a significantly strong correlation is reported ($r>0.67$, $P<0.05$). Although this data does suggest that the timings of these events may be linked to impact tibial shock, the relationships are generally too weak to draw any conclusions across a general population.

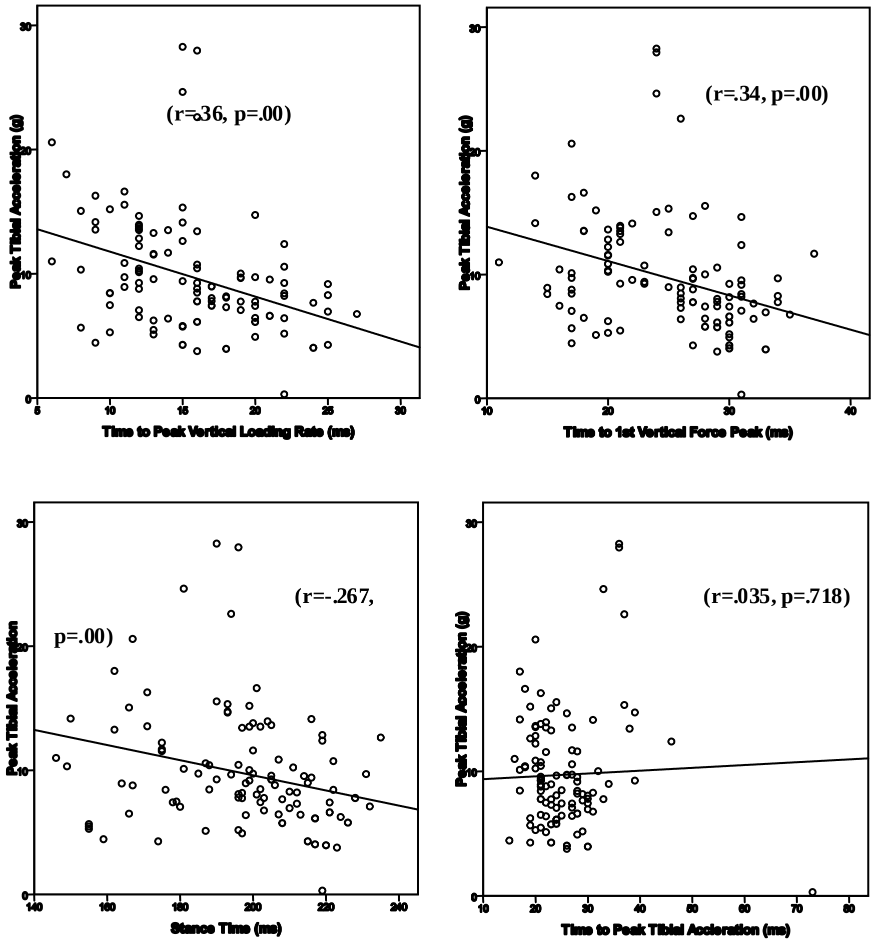


Figure 5.6 Correlation charts for event times compared to peak tibial acceleration

Table 5.2 Correlation values of time of GRF events and tibial acceleration values

Participant		Stance Time	TPVLR	TVFP1	TVFP2D	TPTA
1	Pearson Correlation	.424	-.233	-.728*	-.084	-.501
	Sig. (2-tailed)	.256	.546	.026	.829	.169
2	Pearson Correlation	-.674*	-.481	-.632	-.239	-.385
	Sig. (2-tailed)	.047	.190	.068	.536	.306
3	Pearson Correlation	-.473	-.376	-.622	.028	-.369
	Sig. (2-tailed)	.167	.284	.055	.938	.294
4	Pearson Correlation	-.754*	-.777**	-.828**	.171	-.766**
	Sig. (2-tailed)	.012	.008	.003	.637	.010
5	Pearson Correlation	-.384	.331	-.469	.272	.168
	Sig. (2-tailed)	.348	.424	.241	.515	.691
6	Pearson Correlation	-.847*	-.546	.555	-.921**	-.470
	Sig. (2-tailed)	.016	.205	.196	.003	.287
7	Pearson Correlation	-.163	-.186	-.574	-.158	-.693*
	Sig. (2-tailed)	.675	.632	.106	.684	.038
8	Pearson Correlation	.397	.213	.081	-.023	-.275
	Sig. (2-tailed)	.378	.647	.863	.961	.551
9	Pearson Correlation	-.385	-.561	.092	-.462	-.450
	Sig. (2-tailed)	.346	.148	.829	.250	.263
10	Pearson Correlation	.749*	-.434	-.858**	.009	-.797*
	Sig. (2-tailed)	.020	.243	.003	.981	.010
11	Pearson Correlation	-.764*	-.193	.231	-.850**	-.088
	Sig. (2-tailed)	.027	.647	.581	.008	.835
12	Pearson Correlation	-.486	.066	.400	-.527	.149
	Sig. (2-tailed)	.154	.856	.252	.117	.681
13	Pearson Correlation	.272	-.129	-.605	-.049	-.511
	Sig. (2-tailed)	.446	.723	.064	.894	.131
Overall	Pearson Correlation	-.267**	-.359**	-.336**	-.109	.035
	Sig. (2-tailed)	.004	.000	.000	.251	.718

<u>Terms for Table 5.2</u>

TPVLR = Time to peak vertical loading rate from foot down.

TVFP1 = Time to 1st vertical force peak from foot down.

TVFP2 = Time to 2nd vertical force peak from foot down.

TPTA = Time to peak tibial acceleration from foot down.

***=P<0.05 **=P<0.001**

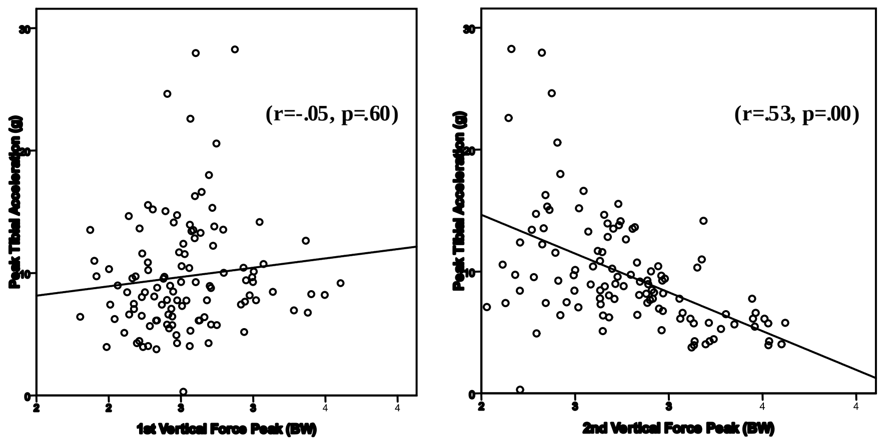


Figure 5.7 Correlation charts for 1st and 2nd vertical force peaks compared to peak tibial acceleration

From the results presented in Table 5.2 there are strong correlations between loading rates and peak tibial accelerations measured by a skin mounted accelerometer for most of the participants' individual data. However for some individuals, many of the peak tibial acceleration values are not significantly correlated to the various GRF variables. In many cases the correlation values are relatively weak ($r<0.5$) and it may be that with an increased amount of trials recorded, more significant and stronger correlations could be identified. Overall, across the data for all participants it would appear that the strongest significant correlation ($r=-0.526$, $P<0.001$) is that of the magnitude of the second force peak (Figure 5.7). This suggests that the lower the

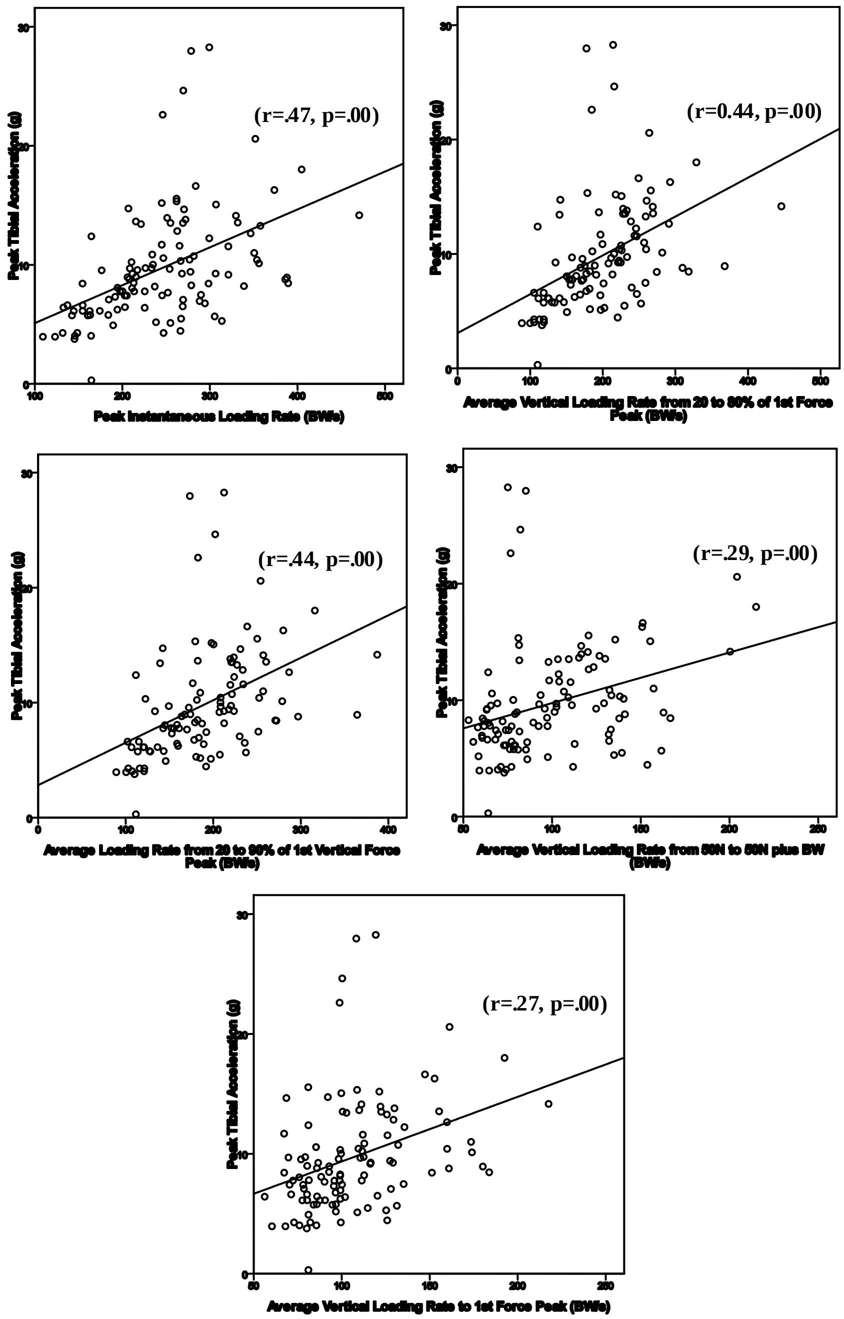


Figure 5.8 Correlation charts for various vertical loading rates compared to peak tibial acceleration

Table 5.3 Correlation values of GRF values and peak tibial acceleration values

Participant	Correlation Value	VFP1	VFP2	PVLR	AVLR	AVL50NT50NBW	AVL20T80	AVL20T90	BFP
1	Pearson Correlation	.707*	-.637	.762*	.731*	.648	.787*	.728*	.191
	Sig. (2-tailed)	.033	.065	.017	.025	.059	.012	.026	.623
2	Pearson Correlation	.104	-.567	.799**	.759*	.705*	.793*	.641	.669*
	Sig. (2-tailed)	.789	.111	.010	.018	.034	.011	.063	.049
3	Pearson Correlation	.596	-.385	.733*	.811**	.591	.809**	.711*	-.251
	Sig. (2-tailed)	.069	.272	.016	.004	.072	.005	.021	.484
4	Pearson Correlation	.319	-.063	.877**	.789**	.662*	.865**	.863**	.492
	Sig. (2-tailed)	.369	.863	.001	.007	.037	.001	.001	.149
5	Pearson Correlation	.549	-.298	.646	.675	.130	.593	.635	.215
	Sig. (2-tailed)	.159	.474	.083	.066	.758	.121	.091	.609
6	Pearson Correlation	-.052	-.332	.483	-.294	.898**	.521	.359	.281
	Sig. (2-tailed)	.912	.467	.272	.522	.006	.230	.430	.541
7	Pearson Correlation	.661	-.112	.740*	.810**	.610	.578	.582	.362
	Sig. (2-tailed)	.053	.774	.023	.008	.081	.103	.100	.339
8	Pearson Correlation	.720	-.661	.681	.615	.123	.502	.452	.345
	Sig. (2-tailed)	.068	.106	.092	.142	.792	.251	.308	.449
9	Pearson Correlation	.217	.213	.746*	-.046	.395	.806*	.779*	.406
	Sig. (2-tailed)	.606	.612	.034	.914	.333	.016	.023	.319
10	Pearson Correlation	.937**	.705*	.915**	.938**	.849**	.887**	.898**	.681*
	Sig. (2-tailed)	.000	.034	.001	.000	.004	.001	.001	.043
11	Pearson Correlation	.775*	.719*	.663	.487	.037	.753*	.739*	.841**
	Sig. (2-tailed)	.024	.045	.073	.220	.930	.031	.036	.009
12	Pearson Correlation	.761*	.628	.407	.350	-.148	.653*	.633*	.838**
	Sig. (2-tailed)	.011	.052	.243	.322	.684	.040	.049	.002
13	Pearson Correlation	.622	-.567	.701*	.638*	.599	.708*	.679*	.192
	Sig. (2-tailed)	.055	.088	.024	.047	.067	.022	.031	.595
Overall	Pearson Correlation	-.051	-.526**	.469**	.274**	.291**	.439**	.439**	.326**
	Sig. (2-tailed)	.595	.000	.000	.004	.002	.000	.000	.000

Terms for Table 5.3

VFP1 = 1st Vertical Force Peak

VFP2 = 2nd Vertical Force Peak

PVLR = Peak Instantaneous Vertical Loading Rate

AVLR = Average Vertical Loading Rate

AVL50NT50NBW = Average Vertical Loading Rate From 50N to 50N Plus BW

AVL20T80 = Average Vertical Loading Rate from 20 To 80% of 1st Vertical Force Peak

AVL20T90 = Average Vertical Loading Rate from 20 To 90% of 1st Vertical Force Peak

BFP = Breaking Force Peak

second GRF peak is, the more likely it is that there is a higher tibial acceleration experienced. This is a surprising result as in previous studies stronger correlations are found between rates of loading and the peak tibial acceleration. The loading rates analysed (Figure 5.8) identified that the strongest correlation ($r=0.469$, $P<0.001$) exists between the peak instantaneous loading rate and the peak tibial accelerations. It would appear the average loading rates from 20 to 80% and 20 to 90% of the increase in force leading up to the 1st impact peak are more effective at identifying the magnitude of tibial accelerations, than taking the average loading rate from initial contact to the 1st force peak. These two methods of analysing the loading rate provide the most amounts of individual significant correlations with 8 and 9 of the participants' data recording significant correlations for the 20-90% and the 20 to 80% average loading rates respectively. The variability in the individual data provides a problem when using force plate data to investigate impact shock. Across the individual data some of the methodologies used to identify detrimental impact shock characteristics are not consistent in the 10 trials of data recorded for each. The 2nd vertical impact peak only has significant positive correlations in any of the individual Pearson's correlations, yet overall has the strongest negative correlation. This highlights the participant variability of such data and suggests that the 2nd force peak may not be the best indicator of peak tibial accelerations.

5.4 Conclusion

It must be acknowledged that skin mounted accelerometers have been questioned in the literature and hold their own inaccuracies when used to measure impact shock in the tibia. However if a rigid coupling is used along with skin stretching techniques, the data recorded gives a good estimation of the shock transmission through the skeletal system. As it was not an option in this study to use invasive rigid attachments to the tibia in order to measure impact shock, the two data collection methods available are using an underfoot force measuring device and a skin mounted accelerometer. The results of this study suggest that using GRF characteristics to predict tibial acceleration is dependent on the individual. With much higher correlations found within individuals compared to across the cohort of participants. The use of a skin mounted accelerometers would seem to be the best practical methodology for comparing impact shock between footwear. The accelerometer can be attached and remain in place between trials when comparing footwear, allowing for a good comparative data to be collected between footwear worn by the each individual participant. With the aims of this research, it would appear to be the most effective way of investigating tibial shock. Although the accelerometer will require equipment to be attached to the participant's body, it will not be affected by force plate targeting as reported in the literature, allowing for more natural movement. Furthermore, it will also provide a portable system that will allow ease of use at an externally located synthetic sports surface. This will allow data to be collected in the environmental conditions typical of field hockey participation, which will report kinetic data that will more accurately describe the injury potential during normal field hockey participation.

Chapter Six

Introduction and evaluation of footwear typically used by field hockey participants

6 Introduction and evaluation of footwear typically used by field hockey participants.

During the following research investigations in this thesis (Chapter 6-9) a set of six commercially available pairs of footwear typically worn by field hockey participants will be tested. The shoes investigated represent different styles typically worn including soccer and running specific footwear. Due to the expense of such shoes the study was limited to two sizes of shoe for each design. Therefore each shoe was sourced in both sizes 9 and 10 (UK shoe sizes) because a short survey of field hockey players at a university revealed these to be the two most common sizes. In this section the footwear is introduced (6.1). The shoes are assigned a descriptive name which they will be referred to throughout this research project to provide continuity, and allow the reader to reference the footwear to the results and discussion more effectively. The footwear is then tested mechanically using a drop-test to measure GRFs variables, recorded from a force plate under the footwear (6.2 to 6.5). The GRF variables investigated in this chapter are the same as the ones used in human locomotion testing of the footwear (Chapter 8).

6.1 The shoes under investigation

Shoe 1: Gryphon Viper

The first shoe used is the **Gryphon Viper** (Figure 6.1) which is a traditional type of hockey shoe manufactured by the Gryphon Hockey Company.

Figure 6.1 Gryphon Viper hockey shoe

The shoe has a set of ‘V’ shaped shallow rubber cleats which appear to be designed to provide linear traction. The cleats are rotated 180 degrees in the heel area compared to the forefoot area. This design appears to be constructed to provide a high co-efficient of friction in the heel during breaking and frictional forces in the opposite direction in the forefoot to propel an athlete forwards. The sides of the shoe are re-enforced to provide protection from side impacts from balls and sticks.

Shoe 2: Gryphon Venom

The second shoe is the **Gryphon Venom** (Figure 6.2) which is a traditional type of hockey shoe manufactured by the Gryphon Hockey Company.

Figure 6.2 Gryphon Venom hockey shoe

The shoe is a more lightweight construction than the Gryphon Viper with a dimpled rubber shallow studded design. This design appears to provide more general traction characteristics with no bias towards any specific directions as seen in the design of the Gryphon Viper.

Shoe 3: Asics Gel Lethal

The third shoe is the **Asics Gel Lethal** (Figure 6.3) which is a traditional type of hockey shoe manufactured by the Asics Company.

Figure 6.3 Asics Gel Lethal hockey shoe

The shoe is lightweight and incorporates a compartment of gel in the heel midsole to provide cushioning properties. The cleat design is deeper than the gryphon shoes which has led to them being banned at some Australian elite venues due to supposed pitch damage during play. The shoe’s design offers less protection than the other hockey specific shoes with no extra protection to the sides of the shoe compared to running or soccer specific shoes.

Shoe 4: Umbro Astroturf

The forth shoe is the **Umbro Astroturf** (Figure 6.4) which is a typical soccer shoe designed specifically for synthetic surfaces which field hockey is played on and is manufactured by the Umbro Company. The shoe has support around the heel area with a soft consistent upper throughout the front part of the shoe designed to allow a soccer player to control and kick a ball.

Figure 6.4 Umbro Astroturf shoe

Shoe 5: Saucony running shoe

The fifth shoe is the **Saucony Running** (Figure 6.5) which is a typical running shoe designed for running on hard surfaces. Many players, including international players have used these types of shoes during competitive play and training. The shoe is manufactured by the Saucony Company.

Figure 6.5 Saucony Running shoe

The outer sole of the Saucony Running shoe does have some very shallow cleats but is designed for running in a forward direction on relatively hard (non-turf) surfaces. Such running shoes have been selected by field hockey participants during play and training even at elite level. As they are not designed for traction on a synthetic turf surface the design may have an effect on the ability of a player to accelerate their centre of mass in a desired direction during a match situation.

Shoe 6: Umbro Moulded

The sixth shoe is the **Umbro Moulded** (Figure 6.6) which is a typical soccer shoe designed specifically for hard turf surfaces and is also used by some players during field hockey on synthetic surfaces. The shoe is manufactured by the Umbro Company. The shoe has a very thin outer sole compared to the other shoes with support at the heel and a soft upper although it does appear to be a tighter fit around the dorsum of the foot compared to the Umbro Astroturf shoes.

Figure 6.6 Umbro moulded shoe

This type of shoe is not allowed on many synthetic sports surfaces although it is still a type of shoe that some field hockey participants choose to wear in practice, particularly on synthetic turf with longer turf fibres.

The differences in the designs of the shoes may affect their performance and the protective function when worn during participation in field hockey based activities. The mass of the shoes which has been reported to have a detrimental effect on endurance, with an effect on energy expenditure 1.9 to 4.7 times that of the same increase in body mass (Holewijn *et al.*, 1992), is reported in Table 6.1.

The size 9 shoes are lighter than the equivalent size 10 in all shoes. The Umbro Astroturf shoes have the largest mass when comparing size 9 shoes, whilst Gryphon Viper have the largest mass in the size 10 versions of the shoes. When comparing the hockey shoes particularly in the size 10s, the Asics Gel Lethal have a much smaller mass: 23% and 18% less than the Gryphon Viper and Venom Respectively. Footwear with a smaller mass has been identified as saving energy and thus increasing endurance (Stefanyshyn and Nigg, 2000). The largest difference is between the Asics Gel Lethal and the Gryphon Viper shoes in their size 10 models (213g). In terms of energy expenditure according to previous research this would be the equivalent of a participant having an extra 405 to 1001g of body mass (Holewijn *et al.*, 1992). Furthermore, research also reported a 1% increase in energy cost with each 0.1kg increase of footwear mass (Jones *et al.*, 1986). This would therefore mean that the Gryphon Viper shoes would increase energy costs by 2% compared to the Asics Gel Lethal, thus increasing the effects of fatigue. During physically demanding sports such as field hockey, an increase in fatigue can have a detrimental effect on skill performance as reported in similar sports (Apriantono *et al.*, 2006;Kellis *et al.*, 2006;Rampinini *et al.*, 2009), as well as increasing the risk of injury (Mizrahi *et al.*, 2000a;Tsai *et al.*, 2009;Mizrahi *et al.*, 2000c;Coventry *et al.*, 2006;Bisiaux and Moretto, 2008;Schlee *et al.*, 2006;Nummela *et al.*, 1996;Derrick *et al.*, 2002). Using a smaller size shoe to decrease the detrimental effects of mass should not be a method to reduce the mass of the footwear. Smaller shoes may have a detrimental effect on injury by increasing in-shoe pressures, particularly in the uppers as previously identified (Chapter 4).

Table 6.1 Mass of footwear investigated

Shoe	Mass of Right Shoe (grams)	Mass of Left Shoe (grams)	Mass of Pair (grams)
Gryphon Viper Size 9	383	389	772
Gryphon Viper Size 10	463	456	919
Gryphon Venom Size 9	361	361	722
Gryphon Venom Size 10	431	433	864
Asics Gel Lethal 9	328	330	658
Asics Gel Lethal 10	348	358	706
Umbro Astroturf Size 9	405	417	822
Umbro Astroturf Size 10	433	448	881
Umbro Moulded Size 9	339	335	674
Umbro Moulded Size 10	362	352	714
Saucony Running Shoe Size 9	376	381	757
Saucony Running Shoe Size 10	393	398	791

The different characteristics of the shoes may affect kinetic data related to injury. Therefore, Chapters 6 to 9 will test the affects of the footwear on impact drop testing, locomotion plantar pressures, locomotion GRFs and locomotion impact shock. Firstly the various shoes are to be mechanically tested for their ability to influence impact forces during a repetitive drop test.

6.2 Material testing of footwear

Sports shoes used for field hockey participation need to be designed for activities involving high frequencies of foot to ground impacts over a sustained period of time. The ability of a sports shoe to reduce the transient forces that have been linked to overuse injury may be related to the ability of the shoe to reduce the peak forces and loading rates during an impact. Previous research has tested a shoe's ability to mechanically attenuate these forces (Aguinaldo and Mahar, 2003;Frederick *et al.*, 1984).

Characteristics of footwear have been measured using a piezoelectric force transducer mounted to a 7.3kg shaft dropped freely onto the forefoot and heel regions of shoes (Frederick *et al.*, 1984). The results of the study suggested that thickness of material influenced impact force attenuation more than the softness of the materials used in in-shoe cushioning systems. Reporting non-linear increases in impact forces for softness and thickness of materials, the study suggests finding the ideal construction of a shoe is problematic. Increases in softness and thickness were found to have a detrimental effect on rearfoot control. The study concluded that the optimum shoe for rear foot control and force attenuation is a thickly soled shoe with a 35 durometer midsole and 15 degree flare. However the results were not tested on human participants. For the best impact force attenuation results mechanically, the results reported what would be expected, that the softest shoe with the thickest midsole produced the smallest impact peak. The non linear effects of thickness and softness of shoes (Figure 6.7) demonstrate that selecting the ideal cushioning that provides enough rearfoot control may not be a simple answer due to the negative effects of softness and thickness on rearfoot motion control as reported in human testing (Robbins and Waked, 1997).

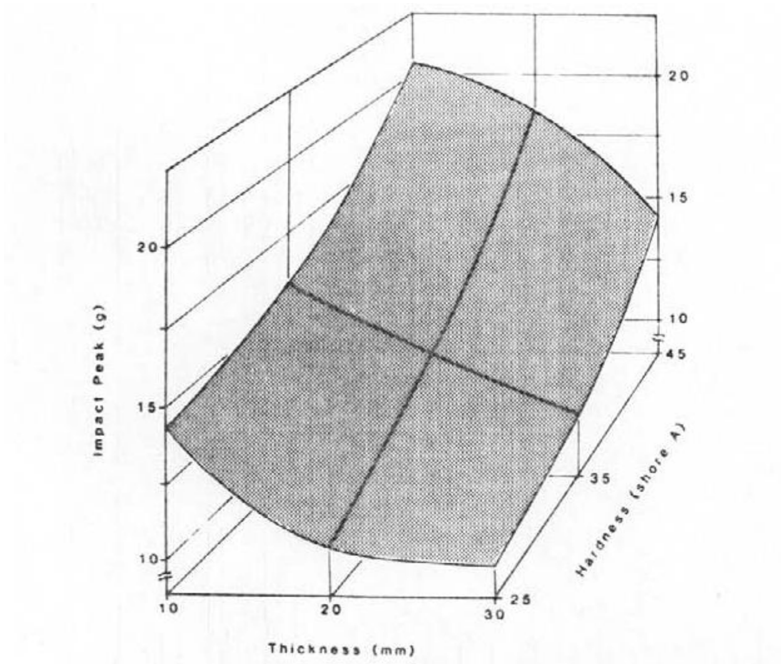


Figure 6.7 Effects of thickness and softness of midsoles in shoes during an impact test, adapted from Frederick (1984)

By dropping an accelerometer attached to a weight into each shoe so the weight would be travelling at 1m.s^{-1} at impact with the shoe, significant differences ($P<0.05$) in the acceleration characteristics (peak and time to peak) of the weight between each of the shoes were reported (McNair and Marshall, 1994). However, during running in the shoes the study reported no significant differences in tibial accelerations between the shoes. A similar study measuring force as opposed to accelerations, used a pendulum to simulate a heel strike (Aerts and De Clercq, 1993). From the results of the study, midsole hardness was found to be related to loading rate. The results also indicated that an increase in deformation of the heel pad area of the shoe (softer) produced a reduction in the loading rate recorded from a force plate. These results agree with a material test of footwear using applied force measurement (with force up to 1400N to simulate 2.5BW of participants) by a hydraulic testing machine (Aguinaldo and Mahar, 2003). This study found that a shoe reported as softer (lower loading rate) during the mechanical test, produced higher loading rates when measuring human locomotion GRFs when compared to another shoe. Similar results were reported from studies investigating the validity of testing sports surfaces (Nigg, 1990). Previous research testing sports surfaces used an impact rig with a 6.8kg spherical head-shaped weight mounted with an accelerometer. The results of the tests found that harder surface peak accelerations were recorded 1ms after impact compared to the softest surface which occurred after 4ms. Magnitudes of accelerations were reported as 300 and 55g for the hard and soft surfaces (Dixon *et al.*, 2000). Further research measuring accelerations during impacts at the heel region of shoes with various cushioning property adjustments, were carried out in such a way that the potential energy at point of impact was between 1.82 and 6.08 J. Furthermore, the results of this same study found that extra cushioning and insoles had more of an effect on shoes with less midsole cushioning and also absorbed a larger proportion of the energy attenuated during low energy impacts (Chiu and Shiang, 2007).

It would appear that a shoe's construction may provide effective attenuation of impact forces during a mechanical test. However, when worn by humans during locomotion the more effective shoes mechanically do not perform in the same manner. The complex nature of the mechanics of the human body may be the reason that identification of force characteristics such as loading rates and force peaks appear to have different levels of influence on the impact shock measured by mounted accelerometers. When considering detrimental impact force characteristics, various studies have found stronger correlations between peak impact tibial accelerations and the average loading rate across various time periods than impact force peaks measured (Bus, 2003;Hennig and Lafortune, 1991;Laughton *et al.*, 2003). A mechanical test should therefore consider the various methodologies for analysing loading rates to test if there is a more effective methodology that identifies characteristics from material tests that are reflected in human locomotion testing.

In order to test the impact cushioning characteristics of the shoes investigated in this research, a mechanical test needs to be performed. By consistently dropping a weight of known mass from a known height into the rear foot area of the shoe, cushioning characteristics of the individual shoes can be measured through use of a force plate. The impact characteristics need to be of a relevant magnitude as to allow comparison between the mechanical test and subsequent analysis of footwear impact kinetics later in this thesis.

6.3 Methodology

A metal pole with a 34mm diameter and a length of 50cm with a mass of 3kg was hung from an indoor winch on the end of a string (Figure 6.8). The shoe being tested was securely attached with adhesive tape to a force platform and a card tube of diameter (50mm) was inserted tightly into the shoe and securely attached to the shoe by adhesive tape. The pole was positioned so it would drop down the card tube with minimum contact. Six shoes were used: Gryphon Viper, Gryphon Venom, Asics Gel Lethal, Umbro Astroturf, Saucony Running, and Umbro Moulded.

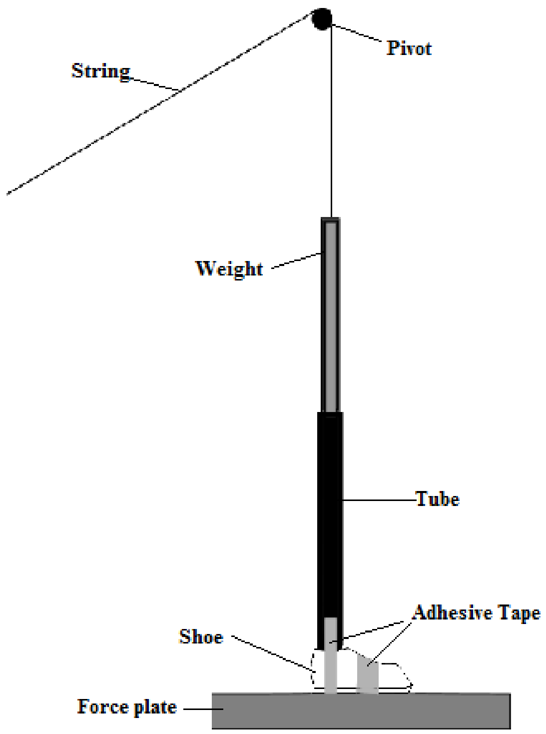


Figure 6.8 Impact force measurement setup

For each shoe the distance between the bottom of the weight and the insole of the shoe was measured and a marker placed on the pole to indicate when the distance was 40cm and 60cm. Vertical GRF data was recorded at 1000Hz while 10 successful drops at each height were performed on each shoe. A rubber mat (8mm thick) was placed over the force plate while 10 further successful drops at both heights were performed using the same procedure as with the footwear tests. The rubber mat was required to protect the platform while providing minimal cushioning to measure impact data using the same test parameters without the cushioning provided from the sports specific footwear. The same GRF variables were calculated as in Chapter 5 using the same Matlab software program.

Given that the drop height is known and without considering air resistance and minimal friction from the string on the winch and the pole on the card, an estimate of the velocity of the pole at contact with the shoes and mat could be calculated.

Using the known equation of motion:

v= final velocity

u=initial velocity

a=acceleration

s=distance

$$v^2 = u^2 + 2as$$

Given that u = 0, we have:

$$v = \sqrt{2as}$$

So for a drop height of 40cm where $a=9.81m.s^{-2}$ and s=0.4m the estimated velocity at impact is

<u>$2.80m.s^{-1}$ to 2d.p.</u>

For a drop height of 60cm where $a=9.81m.s^{-2}$ and s=0.6m the estimated velocity at impact is

<u>$3.43m.s^{-1}$ to 2d.p.</u>

Furthermore the velocity calculation enables the calculation of the kinetic energy (K.E. = m.v^2) and the momentum (Momentum = m.v) of the pole at impact for both conditions (Table 6.2)

Table 6.2 Calculated values of the weight at impact

Drop Height	40cm	60cm
Velocity	2.8 m.s^{-1}	3.4 m.s^{-1}
Kinetic Energy	11.8 J	17.7 J
Momentum	8.4 kg.m.s^{-1}	10.3 kg.m.s^{-1}

This impact test will use methodologies reported in research investigating human locomotion (Hennig and Lafortune, 1991;Nigg *et al.*, 1988;Perry and Lafortune, 1993;Guido *et al.*, 2009;Laughton *et al.*, 2003;Diop *et al.*, 2005;Kong *et al.*, 2009;Pohl *et al.*, 2008), to report time, force peak, and loading rate GRF variables. This is to allow comparisons between this material test and human locomotion tests reporting these variables.

6.4 Results and Discussion

This study has investigated impacts involving relatively large amounts of energy (Table 6.2) compared to previous studies (Chiu and Shiang, 2007;Chi and Schmitt, 2005b). Impact peaks in the hockey and running footwear across both drop heights were between 1281 and 2239N which for a human with a mass of 80kg would be between 1.63 and 2.85BW. These values are comparable to those found in previous mechanical testing research (Aguinaldo and Mahar, 2003), and studies investigating impact peak GRFs during human locomotion (Challis, 2001;McNair and Marshall, 1994;Aguinaldo and Mahar, 2003).

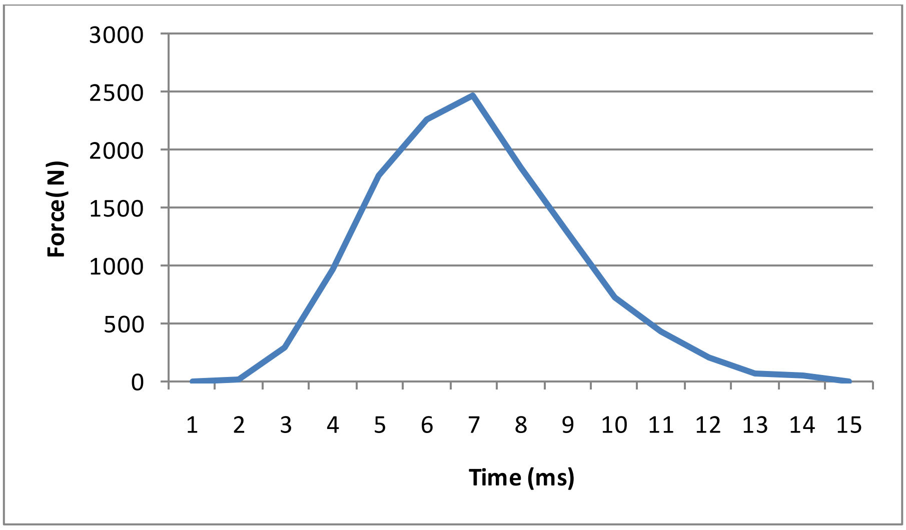


Figure 6.9 Typical impact force data

Typical force data from a single impact drop test shows a single peak occurring in a relatively quick time (Figure 6.9). Mean data for all drop tests are reported in Table 6.3. Figures 6.10 to 6.13 where show the distribution of the data from both drop heights for all conditions. The box plots report the median, the 50% and the 100% distribution of the data (excluding the outliers).

Table 6.4 and Table 6.5 use a method to demonstrate differences in footwear influences on kinetics adapted from previous research on similar effects of footwear (Burnfield *et al.*, 2004).

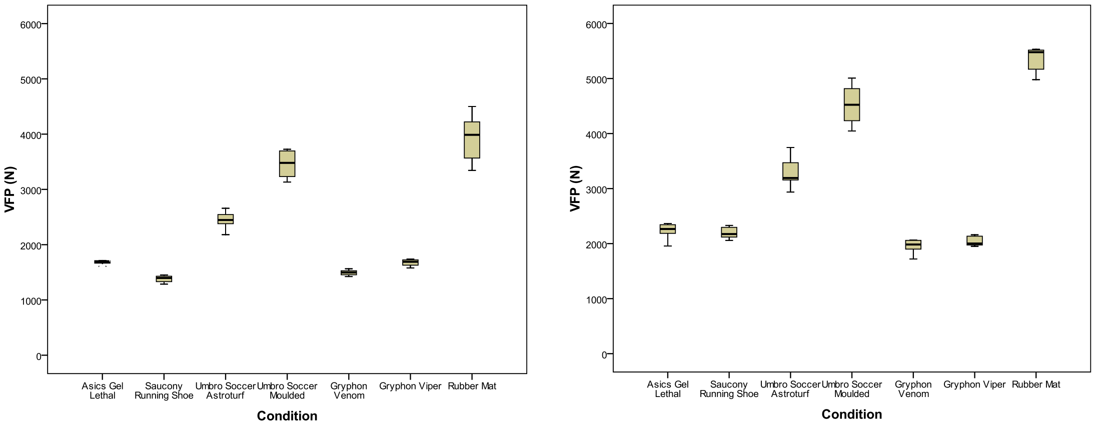
Table 6.3 Mean results of impact drop test

Shoe/Surface	Drop Height (cm)	TPVLR (ms)		TVFPFTD (ms)		VFPN (N)		AVLRN (kN.s-1)		AVL20T90N (kN.s-1)		AVL20T80N (kN.s-1)		PVLR (kN.s-1)	
Asics Gel Lethal	40	3.6	±0.5	7.8	±0.4	1675	±44	215	±15	314	±19	355	±25	415	±17
	60	3.3	±0.5	7.1	±0.6	2119	±281	302	±58	413	±71	455	±88	557	±87
Saucony Running	40	3.5	±0.5	9.0	±0.9	1381	±54	155	±18	190	±12	201	±6	274	±20
	60	3.9	±0.9	8.1	±0.7	2198	±108	272	±34	374	±36	373	±31	425	±56
Umbro Moulded	40	1.9	±0.3	4.2	±0.4	3449	±250	825	±109	1055	±116	1370	±404	1710	±116
	60	1.8	±0.4	3.8	±0.4	4528	±331	1204	±151	1688	±363	1801	±553	2489	±238
Gryphon Viper	40	2.9	±0.8	6.9	±0.8	1676	±59	247	±35	303	±27	339	±31	444	±79
	60	2.0	±0.9	6.0	±0.0	2042	±86	340	±14	387	±44	393	±37	536	±64
Gryphon Venom	40	1.3	±0.5	7.1	±0.7	1494	±48	212	±24	223	±26	232	v18	435	±42
	60	1.6	±0.7	7.1	±0.8	1949	±116	276	±28	318	±26	342	±27	538	±71
Umbro Astroturf	40	2.8	±0.6	5.6	±0.5	2441	±136	440	±58	613	±56	654	±96	847	±95
	60	3.3	±0.9	5.3	±0.5	3290	±267	626	±85	866	±139	889	±117	1162	±193
Rubber Mat	40	2.4	±0.7	3.8	±0.4	3915	±390	1035	±86	1462	±227	1462	±227	1575	±231
	60	2.0	±0.0	3.7	±0.5	5334	232	1480	±227	2355	±452	2486	±419	2595	±348

From the impact characteristics reported (Table 6.3), the time to peak loading rate provided the least valuable results as the values recorded were so small (1 to 3ms in most cases). This meant that through sampling at 1000Hz so data was given to the nearest millisecond, differences in the data may not have been accurately recorded. Therefore, gaining significant level ($p<0.05$) and reliable values were questionable. Furthermore the four other values used to compare the shoes are much more widely reported in literature. The time to peak force took place over a relatively short period across the footwear conditions and drop heights (mean values between 3.8 and 9.0 ms) compared to what has been reported in literature for human running (Hennig and Lafortune, 1991), and in other impact tests (20-30ms) (Frederick *et al.*, 1984). However, due to the mass of the object, the impact peaks were not excessive in the running and hockey specific shoes. The mean times to peak force and times to peak loading rates when compared to peak force through a Pearsons correlation, were found to correlate with values of $r=-0.89$ and $r=-0.28$ respectively. When comparing all five force characteristics across both drop heights they all reported strong significant ($P<0.001$) correlations with $r>0.84$ between each variable except for the time to peak loading rate which reported significant values of r between 0.215 and 0.470. This is important evidence as discussed earlier, impact peaks have been found to be strongly correlated to impact accelerations. Hence a strong correlation between time to force peak, and the magnitude of the force peak provides evidence that the time to peak force is an important value when investigating footwear force attenuation. As there is a very weak correlation when considering the time to peak loading rate, using this methodology may not provide data that should be considered when investigating footwear's ability to attenuate the transient loading forces.

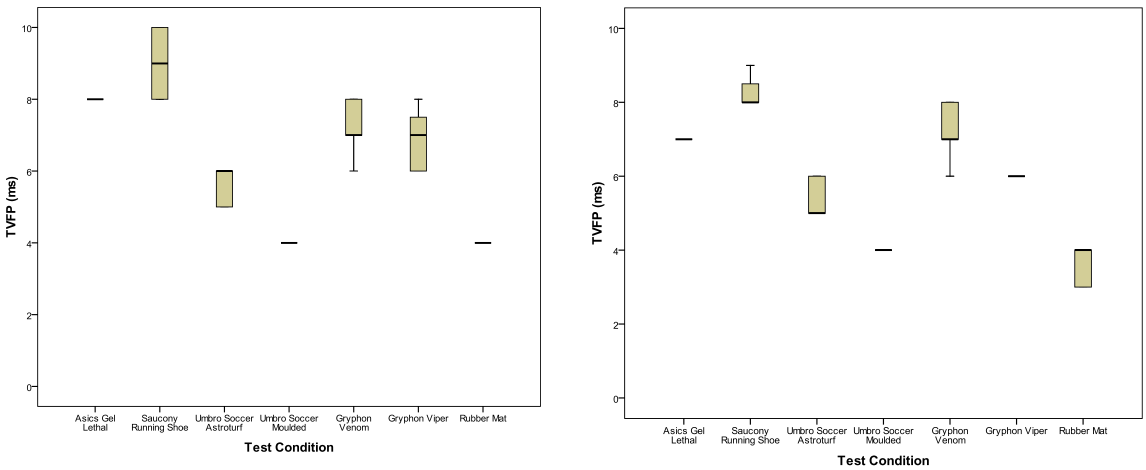
The results of the mechanical test show that on a hard surface (Force Plate) the Saucony Running shoe provided the most positive cushioning characteristics across all the parameters reported for the drop test at 40cm. With significantly ($P<0.05$) higher peak vertical force (Figure 6.10), a longer time to peak vertical impact force (Figure 6.11), a lower average vertical loading rate (Figure 6.12), and a lower peak loading rate than any of the other shoe conditions (Figure 6.13).

Individual tables for effects of footwear-surface on kinetic characteristics linked to overuse injuries are presented (Tables 6.4 and 6.5). Within the tables *=Significant to $P<0.05$, **=Significant to $P<0.001$, ↑=Larger value of the condition being compared across the table compared to the other condition, ↓=smaller value of the condition being compared across the table compared to the other condition, yellow=detrimental cushioning property, and red=beneficial cushioning property. Beneficial or detrimental cushioning properties were highlighted for values that have been linked to overuse injury impacts in human locomotion. Beneficial characteristics were: a longer TPVLR and a longer TVFP1 which have both been identified earlier in this thesis as being correlated to a reduction in tibial accelerations (Table 5.2). A significantly smaller AVLR and PVLR were also considered beneficial as both had also been found to be correlated to a reduction in tibial accelerations (Table 5.3). Detrimental characteristics were therefore reported for shorter TPVLR and TVFP1, while larger AVLP and PVLR were also considered detrimental. The VFP1 is also reported in these tables as it is reported as an area of interest in previous literature.



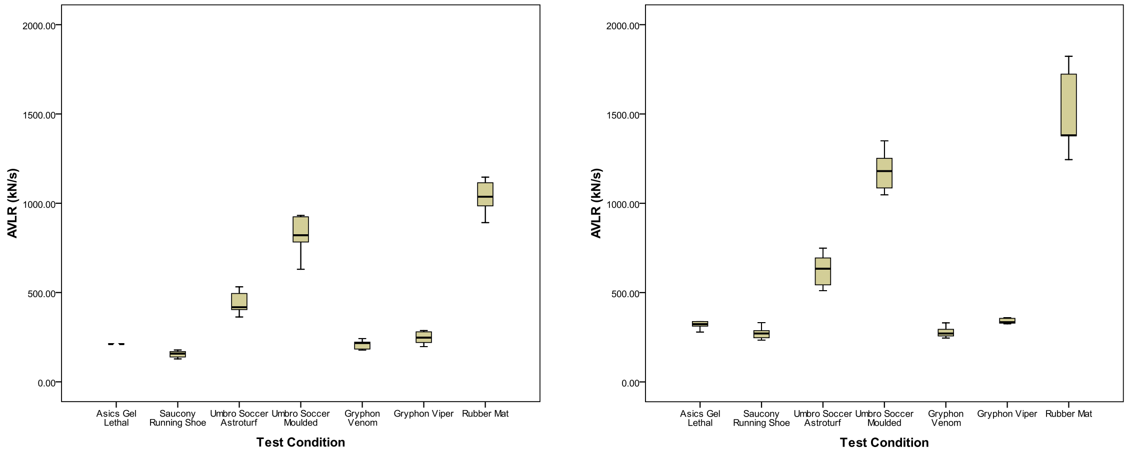
a)40cm b)60cm

Figure 6.10 Distribution of vertical force peak (VFP) values for all drop conditions from a height of a) 40cm and b) 60cm



a)40cm b)60cm

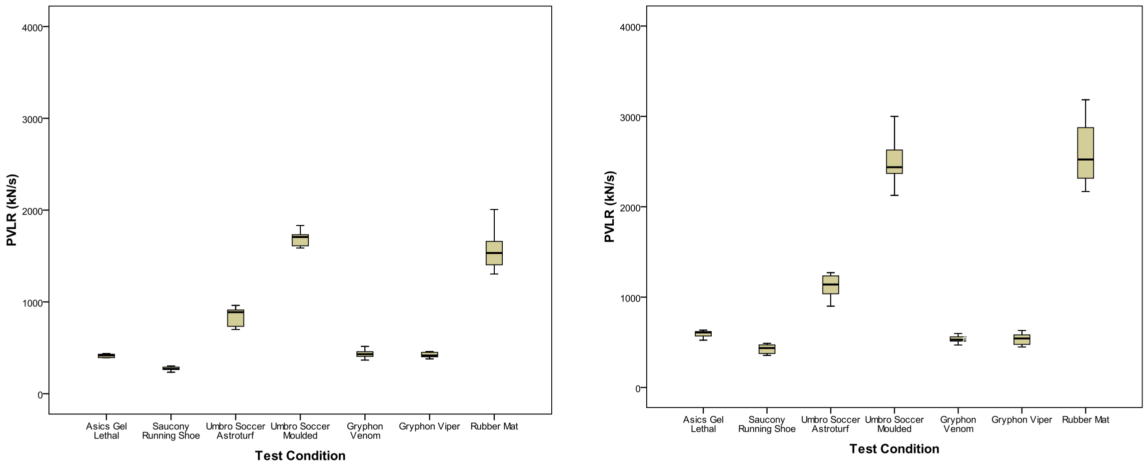
Figure 6.11 Distribution of time from contact to vertical force peak (TVFP) values for all drop conditions from a height of a) 40cm and b) 60cm



a)40cm

b)60cm

Figure 6.12 Distribution of average loading rate (AVLR) values for all drop conditions from a height of a) 40cm and b) 60cm



a)40cm

b)60cm

Figure 6.13 Distribution of vertical loading rate (PVLR) values for all drop conditions from a height of a) 40cm and b) 60cm

However the magnitude of the VFP1 is reported as influencing the impact transient magnitude only moderately compared to the other variables in previous literature using bone mounted accelerometers (Hennig *et al.*, 1993). Reported as occurring later in the stance phase (Whittle 1999), the relationship between the skin mounted accelerometer and VFP1 measurements in this thesis reported no significant correlation (Table 5.3, r=-0.051, P=0.595). Therefore for the remainder of this thesis, while being identified as a variable of interest, its magnitude will not be considered as influencing the kinetics linked to injury during running ($5m.s^{-1}$) and jogging ($3.3m.s^{-1}$).

The Gryphon Venom and Gryphon Viper had significantly smaller mean peak forces than the Saucony Running during the 60cm drop tests. However, due to the mean time to peak force being longer in the Saucony Running, the mean average loading rates were significantly lower compared to the Gryphon Viper and Gryphon Venom. The mean peak loading rate was found to be higher in the field hockey specific footwear. This was a similar trend found in the hockey-specific shoes compared to the Saucony Running. The Gryphon Venom and Asics Gel Lethal reported mean force peaks within 2% of the Saucony Running yet for the mean average loading rate and the mean peak loading rate, the hockey shoes were all found to be of a magnitude >20% compared to the Saucony Running Shoes. With high correlations reported in the literature between loading rates and detrimental tibial impact accelerations (Hennig *et al.*, 1993), it would suggest that the Saucony Running Shoes are more effective in cushioning the impact across both drop heights compared to all the other shoes. The average loading rate also provides similar results with a lower rate of loading recorded for the Saucony Running compared to all the other shoes except the Gryphon Venom. It would appear from the data however that the Umbro Astroturf shoes do not offer as much cushioning as the running and hockey shoes. With increases of over 40% across all force characteristics reported at each drop height, the shoe provides less cushioning than the other non soccer specific shoes, in particular compared to the Saucony Running Shoes with a mean peak loading rate over 3 times as large. It is clear from Tables 6.4f and 6.5f that across the mean force characteristics reported, the Umbro Astroturf shoes provide less cushioning than the running and hockey specific footwear. As may be expected due to their stiff deep cleat design, the Umbro Moulded boots provided the least favourable cushioning characteristics out of all the footwear conditions (Tables 6.4c and 6.5c).

Table 6.4 (a-g) Comparison of shoe impact conditions for the 40cm drop height

a) Asics Gel Lethal

Condition compared to Asics Gel Lethal	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Saucony Running	↑	↓*	↑**	↑**	↑**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Viper	↑*	↑**	↓**	↓*	↓
Gryphon Venom	↑**	↑*	↑**	↑	↓
Umbro Astroturf	↑**	↑*	↓**	↓**	↓**
Rubber Mat	↑**	↑**	↓**	↓**	↓**

b) Saucony Running Shoe

Condition compared to Saucony Running	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓	↑*	↓**	↓**	↓**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Viper	↑**	↑**	↓**	↓**	↓**
Gryphon Venom	↑**	↑**	↓**	↓**	↓**
Umbro Astroturf	↑*	↑**	↓**	↓**	↓**
Rubber Mat	↑*	↑**	↓**	↓**	↓**

c) Umbro Moulded

Condition compared to Umbro Moulded	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓**	↑**	↑**	↑**
Saucony Running	↓**	↓**	↑**	↑**	↑**
Gryphon Viper	↓**	↓**	↑**	↑**	↑**
Gryphon Venom	↑*	↓**	↑**	↑**	↑**
Umbro Astroturf	↓*	↓**	↑**	↑**	↑**
Rubber Mat	↓	↑**	↓**	↓**	↑**

d) Gryphon Viper

Condition compared to Gryphon Viper	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓*	↓**	↑	↑*	↑
Saucony Running	↓**	↓**	↑**	↑**	↑**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Venom	↑**	↓*	↑**	↑*	↑
Umbro Astroturf	↑	↑**	↓**	↓**	↓**
Rubber Mat	↑*	↑**	↓**	↓**	↓**

e) Gryphon Venom

Condition compared to Gryphon Venom	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓*	↓**	↓	↑
Saucony Running	↓**	↓**	↑**	↑**	↑**
Umbro Moulded	↓*	↑**	↓**	↓**	↓**
Gryphon Viper	↓**	↑*	↓**	↓*	↑
Umbro Astroturf	↓**	↑**	↓**	↓**	↓**
Rubber Mat	↓**	↑**	↓**	↓**	↓**

f) Umbro Soccer Astroturf

Condition compared to Umbro Astroturf	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓*	↑**	↑**	↑**
Saucony Running	↓*	↓**	↑**	↑**	↑**
Umbro Moulded	↑*	↑**	↓**	↓**	↓**
Gryphon Viper	↓	↓**	↑**	↑**	↑**
Gryphon Venom	↑**	↓**	↑**	↑**	↑**
Rubber Mat	↑	↑**	↓**	↓**	↓**

g) **Rubber Mat**

Condition Compared To Rubber Mat	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓**	↑**	↑**	↑**
Saucony Running	↓*	↓**	↑**	↑**	↑**
Umbro Moulded	↑	↓*	↑**	↑**	↓**
Gryphon Viper	↓*	↓**	↑**	↑**	↑**
Gryphon Venom	↑**	↓**	↑**	↑**	↑**
Umbro Astroturf	↓	↓**	↑**	↑**	↑**

Table 6.5 (a-g) Comparison of shoe impact conditions for the 60cm drop height

a) **Asics Gel lethal**

Condition compared to Asics Gel Lethal	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Saucony Running	↓**	↓**	↑	↑**	↑**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Viper	↑**	↑*	↑*	↓	↑*
Gryphon Venom	↑**	↓	↑**	↑*	↑
Umbro Astroturf	↓	↑**	↓**	↓**	↓**
Rubber Mat	↑**	↑**	↓**	↓**	↓**

b) **Saucony Running Shoe**

Condition Compared To Saucony Running Shoe	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↑**	↑**	↓	↓**	↓**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Viper	↑**	↑**	↑*	↓**	↓**
Gryphon Venom	↑**	↑**	↑**	↓	↓**
Umbro Astroturf	↑**	↑**	↓**	↓**	↓**
Rubber Mat	↑**	↑**	↓**	↓**	↓**

c) **Umbro Soccer Moulded**

Condition compared to Umbro Moulded	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓**	↑**	↑**	↑**
Saucony Running	↓**	↓**	↑**	↑**	↑**
Gryphon Viper	↓	↓**	↑**	↑**	↑**
Gryphon Venom	↑	↓**	↑**	↑**	↑**
Umbro Astroturf	↓**	↓**	↑**	↑**	↑**
Rubber Mat	↓	↑	↓**	↓**	↓

d) Gryphon Viper

Condition Compared To Gryphon Viper	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓*	↓*	↑	↓*
Saucony Running	↓**	↓**	↓**	↑**	↑**
Umbro Moulded	↑	↑**	↓**	↓**	↓**
Gryphon Venom	↑	↓*	↑	↑**	↓
Umbro Astroturf	↓*	↑**	↓**	↓**	↓**
Rubber Mat	=	↑**	↓**	↓**	↓**

e) Gryphon Venom

Condition Compared To Gryphon Venom	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↑	↓**	↓*	↓
Saucony Running	↓**	↓**	↓**	↑	↑**
Umbro Moulded	↓	↑**	↓**	↓**	↓**
Gryphon Viper	↓	↑*	↓	↓**	↑
Umbro Astroturf	↓**	↑**	↓**	↓**	↓**
Rubber Mat	↓	↑**	↓**	↓**	↓**

f) Umbro Astroturf

Condition Compared To Umbro Astroturf	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↑	↓**	↑**	↑**	↑**
Saucony Running	↓**	↓**	↑**	↑**	↑**
Umbro Moulded	↑**	↑**	↓**	↓**	↓**
Gryphon Viper	↑*	↓**	↑**	↑**	↑**
Gryphon Venom	↓**	↓**	↑**	↑**	↑**
Rubber Mat	↑**	↑**	↓**	↓**	↓**

g) Condition compared to Rubber Mat

Condition Compared To Rubber Mat	TPVLR (ms)	TVFP1 (ms)	VFP1 (N)	AVLR (N/s)	PVLR (N/s)
Asics Gel Lethal	↓**	↓**	↑**	↑**	↑**
Saucony Running	↓**	↓**	↑**	↑**	↑**
Umbro Moulded	↑	↓	↑**	↑*	↑
Gryphon Viper	=	↓**	↑**	↑**	↑**
Gryphon Venom	↑	↓**	↑**	↑**	↑**
Umbro Astroturf	↓**	↓**	↑**	↑**	↑**

It is important to recognise that all the shoes except for the Saucony Running Shoes are constructed for use on a synthetic sports surface where the cleat designs would sink into the soft synthetic cloth top layer to provide traction and also support the shoes out sole at the base of the cleats. So this material test may provide different results if the surface was softer and allowed the shoes' cleats to sink into the surface more. Synthetic surfaces however do vary with water based and sand based surfaces providing very different characteristics so when considering each shoe the surface to which the shoe will be used on will be a factor. The effect of a very rigid deep cleat is seen from the results for the Umbro Moulded shoes. The shoes' rigid cleat and lack of cushioning midsole provide very little cushioning with mean peak loading rates over 4 times that of some of the other shoes and a mean peak vertical force over double that of the running and hockey shoes for the 60cm drop test. For the 40cm drop test the mean peak loading rate for the Umbro Moulded is larger in magnitude than the Rubber Mat condition tested at the same height.

6.5 Conclusion

Clearly from these mechanical test results, the soccer specific shoes provide the least cushioning characteristics out of the shoes and it would appear on a hard surface that the Saucony Running Shoes provide the best. Previous studies have shown however that mechanical tests do not always reflect the outcomes of impact tests for shoes worn during human locomotion (McNair and Marshall, 1994;Aguinaldo and Mahar, 2003). The data from this mechanical test will provide valuable information when assessing similar variables recorded during human movement in the following chapters.

Chapter Seven

Effects of Footwear on In-Shoe Peak Plantar Pressures

Aspects of this work were presented at the Staffordshire Conference on Clinical Biomechanics – 2008, Stoke-on-Trent

7 Effects of Footwear on In-Shoe Peak Plantar Pressures

This study investigated the differences between the peak plantar pressures experienced during human locomotion between footwear typically worn during field hockey participation.

7.1 Introduction

During human locomotive movement the forces applied to the plantar region of the feet are not evenly distributed. It has been reported that the most heavily loaded part of the forefoot is the first ray with about 29% of body weight acting under the metatarsal head (Jacob, 2001). Areas of localised pressure on the plantar surface of the foot, can lead to detrimental effects such as stress fractures of the metatarsal bones and foot ulceration (Cavanagh, 2004). One of the major goals of any footwear intervention must be to protect the foot at sites that are at risk of skin abaisive injuries by reducing pressure to a low level. Previous literature reported specific loading characteristics during specific sports movements (Eils *et al.*, 2004). Therefore research conducted on human participants must recreate the specific movements under investigation. Using an in-shoe pressure measuring system during locomotion, localised plantar pressure can be calculated over specific areas of the foot to identify the location and magnitude of peak pressures. In a study investigating plantar pressure distribution in fencing specific shoes and court shoes, it was found that court shoes reduced peak pressures during a fencing lunge (Geil, 2002). Similarly, significant differences were found between two pairs of running shoes (training and racing flats) for peak pressure and peak force across various areas of the plantar region of the foot during running at self selected speeds along a 10m runway (Wiegerinck *et al.*, 2009). The results of these investigations would suggest that sports-specific footwear can influence plantar pressure distribution. If footwear used during field hockey participation which distributed localised pressure effectively to reduce peak pressures could be identified, this would facilitate the reduction of peak pressures being applied to the plantar region during participation. With the high impact characteristics sustained over time during field hockey participation including matches, training and warming up, there is a need for field hockey footwear to be designed to consider the distribution of the pressure experienced at the plantar region of the foot. There is currently a paucity of information regarding the pressure distribution during locomotion for footwear typically used in field hockey participation. Identifying footwear that reduces peak pressures across a population will provide useful information for field hockey participants in choosing the most suitable footwear. This is particularly valuable for those participants who have suffered previous injuries linked to excessive peak pressures and thus are at a higher risk of injury (Merza and Tesfaye, 2003;Hootman *et al.*, 2002).

7.2 Methodology

Eight healthy males (Age 23 $\pm$ 3.89 years, Height 170 $\pm$ 8.12 cm and Mass 68.88 $\pm$ 10.16 kg) were recruited from a student population. All participants gave informed consent approved by the University Ethics Review Board. Plantar Pressure data collection was performed in a biomechanics laboratory with a temporary synthetic sports turf surface in place (Figure 7.1 and Figure 7.2c). This surface provided a more closely matched alternative to an outdoor hockey specific surface than the hard laboratory floor. As this study is investigating all levels of hockey participation it should be noted that synthetic surfaces in the UK are of varying standards with many severely worn, providing less cushioning. This in-lab alternative therefore, may be more typical of many synthetic surfaces on which hockey participation takes place than a newly laid, international standard surface. In-shoe pressure data was collected on the plantar region of their non-dominant foot using a *Tekscan*® in-shoe pressure measurement system. Only one foot was analysed for each footwear condition. This was considered acceptable as previous research found that significant differences were only reported between dominant and non dominant GRFs and peak pressures in dynamic sideways movements (Munro *et al.*, 1987;Guldemond *et al.*, 2007a;Wong *et al.*, 2007). The in-shoe sensors were checked in between each change of footwear and carefully fitted to cover the plantar region of the foot. A single calibration was performed to the manufacturer's guidance at the start of the data collection session and the same sensor was used for each participant's data collection trials in all the types of footwear. This methodology of using the same sensor and a single calibration has been shown to improve accuracy and reliability of the system (Quesada *et al.*, 1997). As this study is interested in comparing the effects of each footwear condition within each participant across a population, this methodology of keeping the same sensor and calibration provides an effective comparison of the footwear conditions.

Once the in-shoe sensor (F-Scan®) and cuffs were attached (Figure 7.2a-b) the participants jogged and ran 18m across an artificial surface at speeds of 3.33 $m.s^{-1}$ and $5 m.s^{-1}$ respectively in six different pairs of sports specific footwear (Figure 6.1-6.6). The 18m runway was positioned between a set of timing gates spaced 6 metres apart with 6m to speed up and 6m to slowdown either side of the timing gates (Figure 7.1). The runway length was longer than that used in similar previous research (Wiegerinck *et al.*, 2009;Aguinaldo and Mahar, 2003), which allowed participants to accelerate up to the desired speeds and run at a consistent speed before slowing down. A crash mat positioned against the wall at the end of the runway allowed participants to maintain a constant speed through the timing gates with a reduced risk of injury.

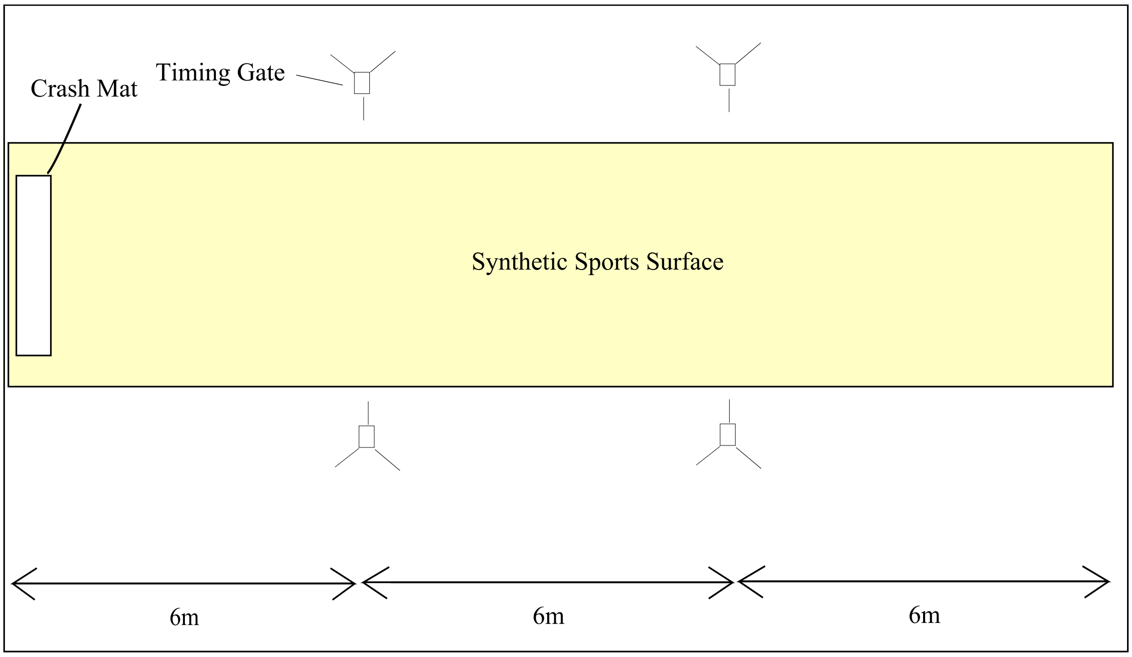


Figure 7.1 Diagram of the laboratory setup

The participants repeated three trials at each speed wearing the six different sports specific footwear in a randomised order. Across the three trials recorded for each footwear condition, nine periods of foot to ground contact were analysed. Participants were given regular rest periods in between trials (30 secs) and between footwear conditions (5 minutes) to restrict the onset of fatigue which has been shown to influence plantar pressure data (Weist *et al.*, 2004). Recording nine of the performed foot to ground impacts for eight participants is in line with the recommended trials per participants in such biomechanical studies (Bates *et al.*, 1992).

a) Participant with F-Scan sensor inserted into shoe

b) F-Scan Sensor

c) Set up of the runway

Figure 7.2 Equipment and laboratory setup. (a) Participant with F-Scan sensor inserted into shoe, (b) F-Scan sensor, and (c) Runway.

7.2.1 Data processing and statistics

Using the Tekscan software, each footwear condition's pressure patterns were evaluated to manually identify the plantar regions of the 1st metatarsal head, the 2nd and 3rd metatarsal head, the 4th and 5th metatarsal head, and the calcaneum as shown in Figure 7.3. Peak pressures in each area of the plantar region were exported for the selected stance phases. Using Excel (Microsoft, USA) spread sheets, the peak pressure values were identified for each plantar region of the foot selected. AVONA's were run to compare the effects of the different footwear being tested.

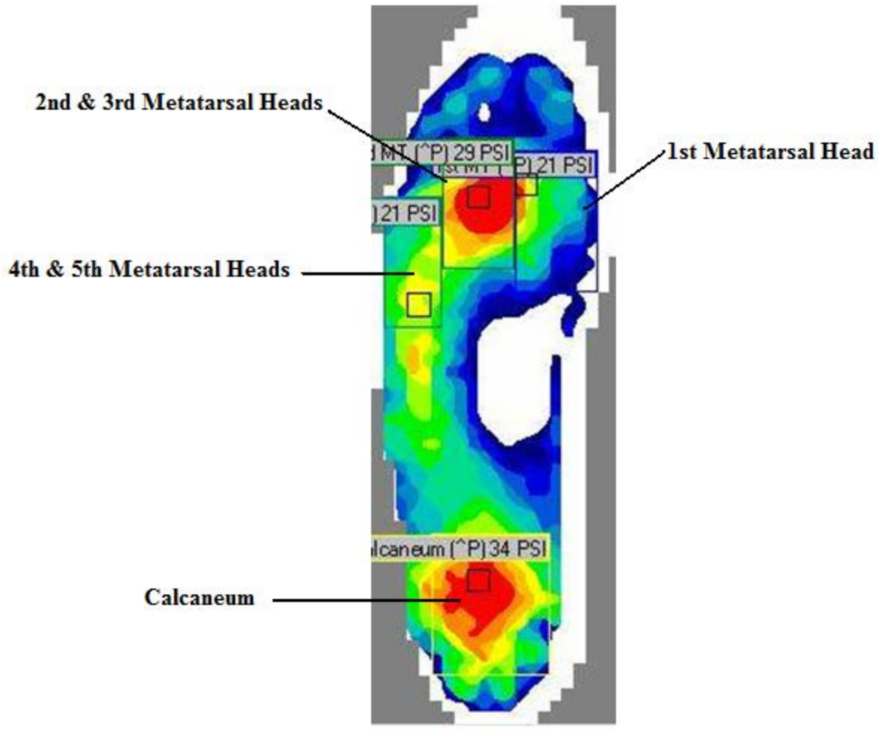


Figure 7.3 Screen shot of the areas of the plantar region identified by the oblong areas highlighting (in clockwise from the top left) the 4th and 5th metatarsal heads, the 2nd and 3rd metatarsal heads, and the calcaneum plantar regions.

7.3 Results and discussion

The mean peak pressures for all participants are approximately in the same range as those reported for shod locomotion in previous research (Burnfield *et al.*, 2004;Tessutti *et al.*, 2010;Guldemond *et al.*, 2007a). The results shown in Table 7.1 have relatively large standard deviations providing evidence of large variability in the data.

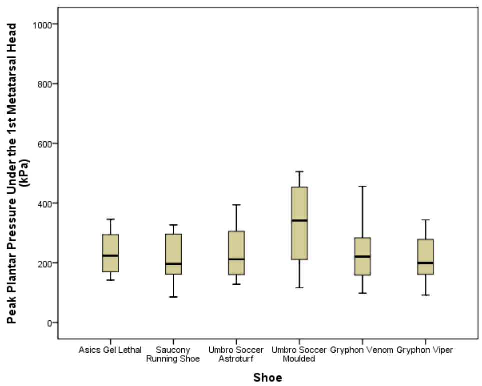
Running compared to jogging produced significantly higher peak pressures (P<0.05) at the 2nd and 3rd metatarsal heads and over the entire plantar region as a whole. No significant (P<0.05) differences between shoes were identified when comparing the mean peak pressure values across all participants were identified. A value of P=0.074 was reported when comparing the Gryphon Viper to the Umbro Moulded, an increase in the population size of this study may have produced a significant difference at the p<0.05 level across the population. The variability in the data is shown across the box plots in Figure 7.4, which show the median values, the 50% data and the 100% data range (excluding outliers). The large variability of the data highlights that an increased population size may be required to identify significant differences. Although in general, the data in Figure 7.4 does suggest that peak pressures across the population in this research do not demonstrate many particular trends. However, it does appear that for the plantar surface under the first metatarsal, the Umbro Moulded shoes expose the participants to larger peak pressures. The peak pressures for this area of the foot in the Umbro Moulded footwear are exposed to the largest peak pressure compared to the other specific plantar region areas. This agrees with previous research investigating this area of the foot (Jacob, 2001).

Significant differences were not reported when comparing the means for the population in this study. However, by performing individual evaluations for each participant, differences can be investigated for the influence of footwear on the peak pressures recorded in individual participants. Performing ANOVAs comparing the peak pressures in the plantar region of the foot between the shoes for individual participants identified significant differences ($P<0.05$) for many of the participants. Figures 7.5 to 7.9 show how many of the eight participants reported significant higher and lower peak pressures within each footwear condition, compared to any of the other five footwear choices. The general trend from this data suggests that footwear with the deepest cleats were found to have significant increases in peak pressures in most of the areas of the foot. The Umbro Moulded shoes on average reported the most number of participants with significantly larger peak pressures. However the same shoes also reported favourable results compared to other shoes for peak pressures under the 2^{nd} and 3^{rd} MTHs. It would appear from this data that most of the participants are loading the other areas of the foot in the Umbro Moulded footwear.

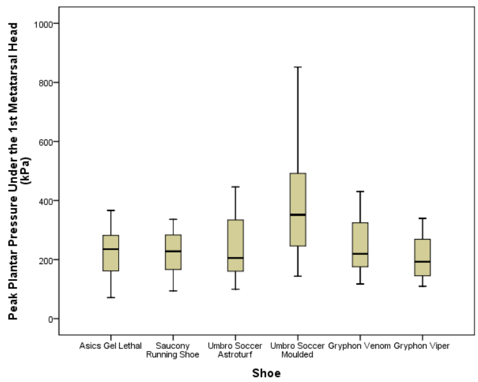
The results of the within participant analysis of the footwear demonstrates how footwear may be prescribed to reduce the chance of a specific injury. If a patient had suffered previous 2^{nd} or 3^{rd} metatarsal stress fractures the Umbro Moulded may be an effective shoe for reducing the onset of future injuries. However this would not be ideal as the other areas of the foot may then be subjected to excessive peak pressures resulting in injury.

Table 7.1 Mean Peak Localised Pressures for all Participants

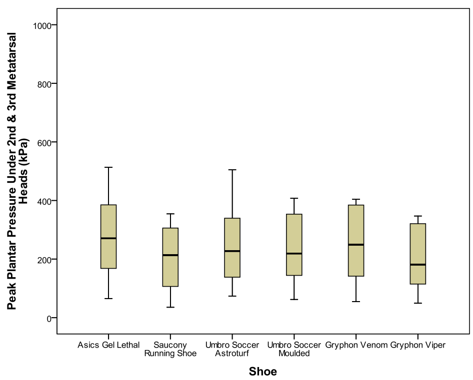
Condition		1st MTH (kPa)		2nd & 3rd MTH (kPa)		4th & 5th MTH (kPa)		Calcaneum (kPa)		Entire Plantar surface (kPa)	
Asics Gel Lethal	Jogging	252	±117	278	±146	180	±125	240	±139	300	±127
	Running	224	±93	279	±127	177	±91	239	±104	312	±108
Saucony Running	Jogging	214	±82	204	±116	150	±90	190	±86	241	±92
	Running	208	±79	230	±138	167	±102	192	±120	252	±115
Umbro Astroturf	Jogging	215	±84	205	±119	152	±119	198	±115	255	±109
	Running	223	±81	214	±118	145	±89	231	±146	282	±132
Umbro Moulded	Jogging	234	±92	249	±145	159	±99	254	±151	289	±135
	Running	243	±119	266	±178	167	±121	239	±110	299	±155
Gryphon Venom	Jogging	329	±141	238	±125	192	±101	203	±115	344	±122
	Running	397	±224	292	±203	207	±111	178	±90	415	±207
Gryphon Viper	Jogging	235	±113	251	±137	175	±117	218	±121	277	±119
	Running	248	±107	282	±163	171	±94	245	±171	315	±157



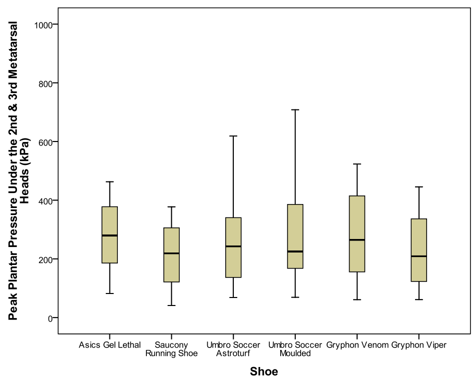
a) Jogging (3.3m.s^{-1})



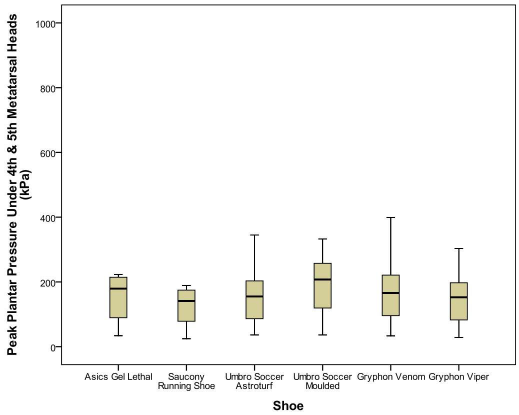
b) Running (5m.s^{-1})



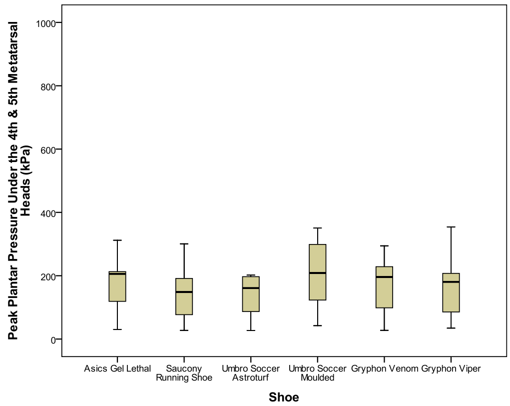
a) Jogging (3.3m.s^{-1})



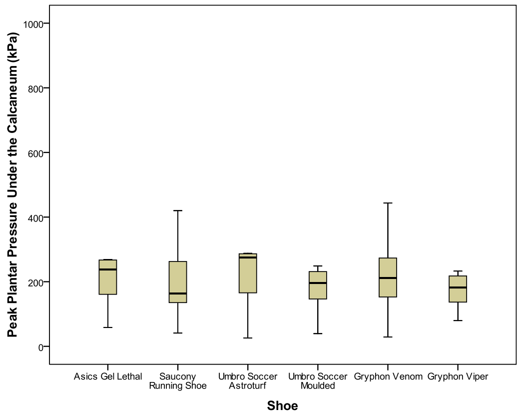
b) Running (5m.s^{-1})



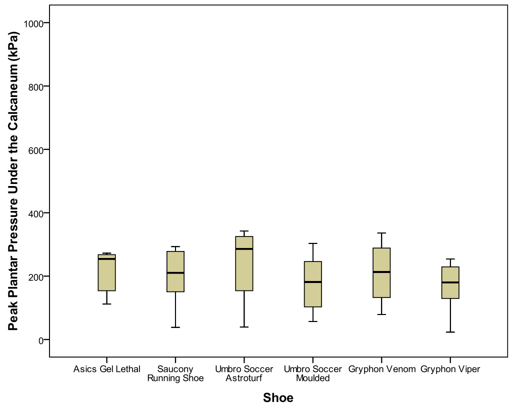
a) Jogging (3.3m.s^{-1})



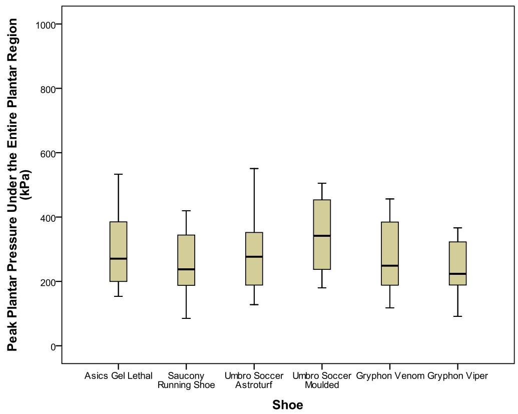
b) Running (5m.s^{-1})



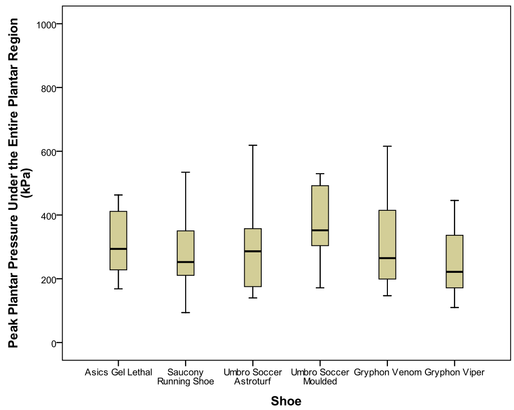
a) Jogging (3.3m.s^{-1})



b) Running (5m.s^{-1})

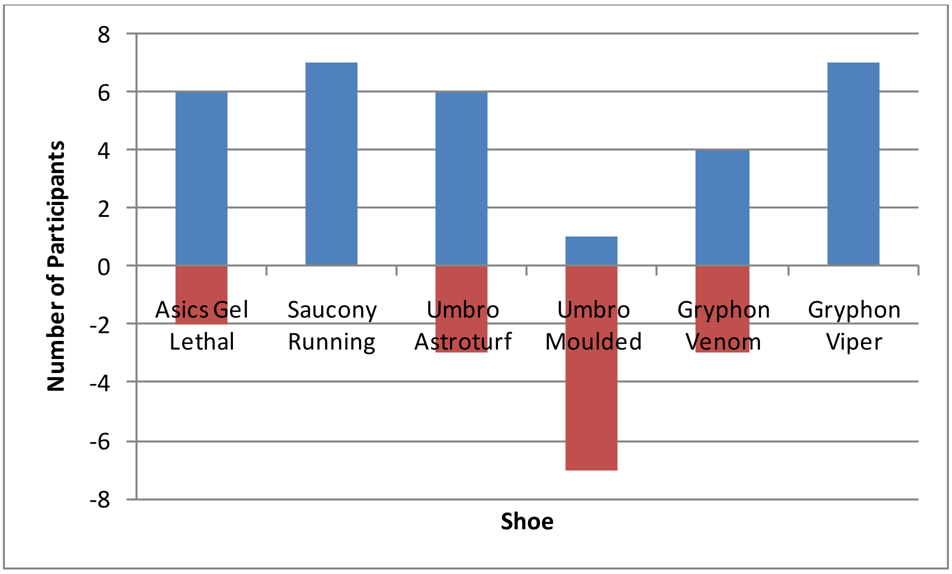


a) Jogging (3.3m.s^{-1})

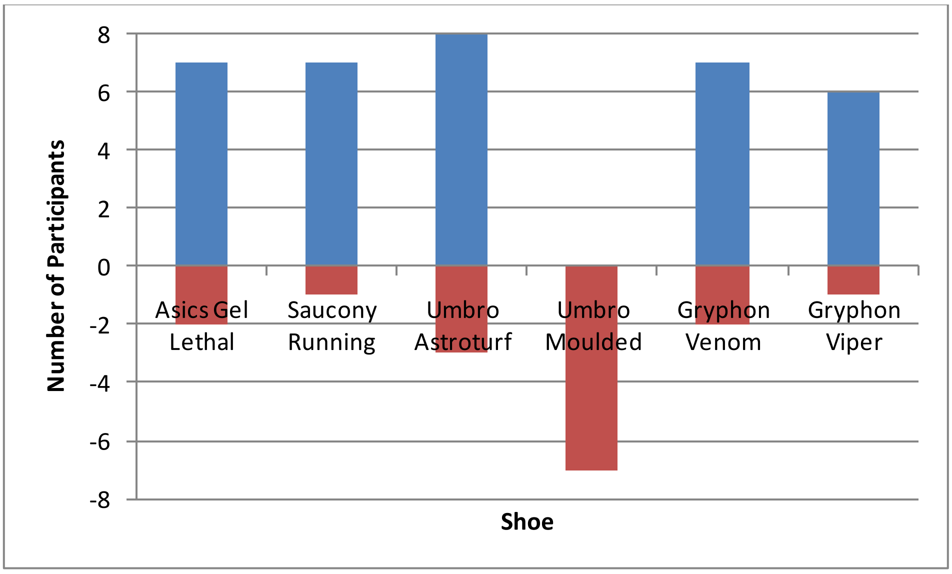


b) Running (5m.s^{-1})

Figure 7.4 Distribution of peak pressure recorded during (a) Jogging (3.3m.s^{-1}) (b) Running (5m.s^{-1})



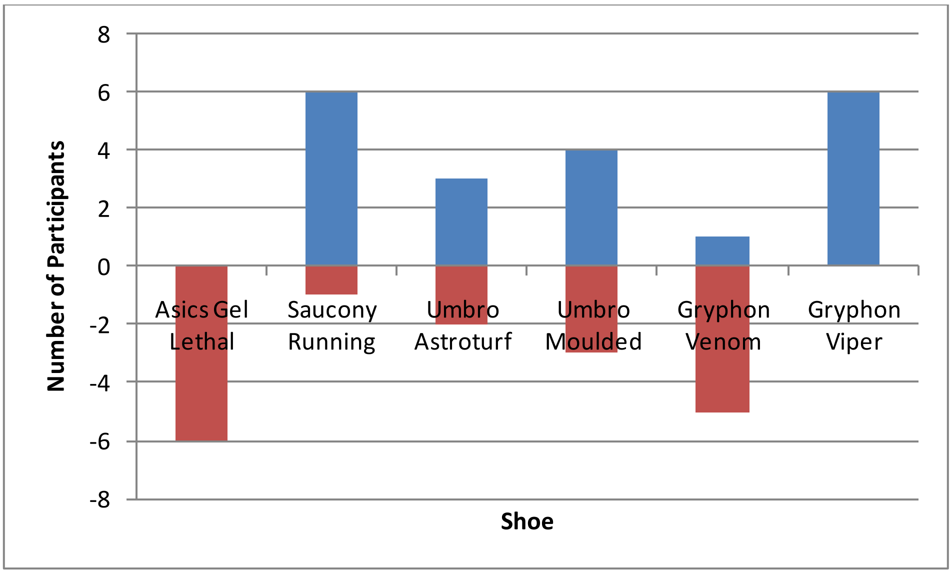
a) Plantar region of 1st metatarsal head during jogging



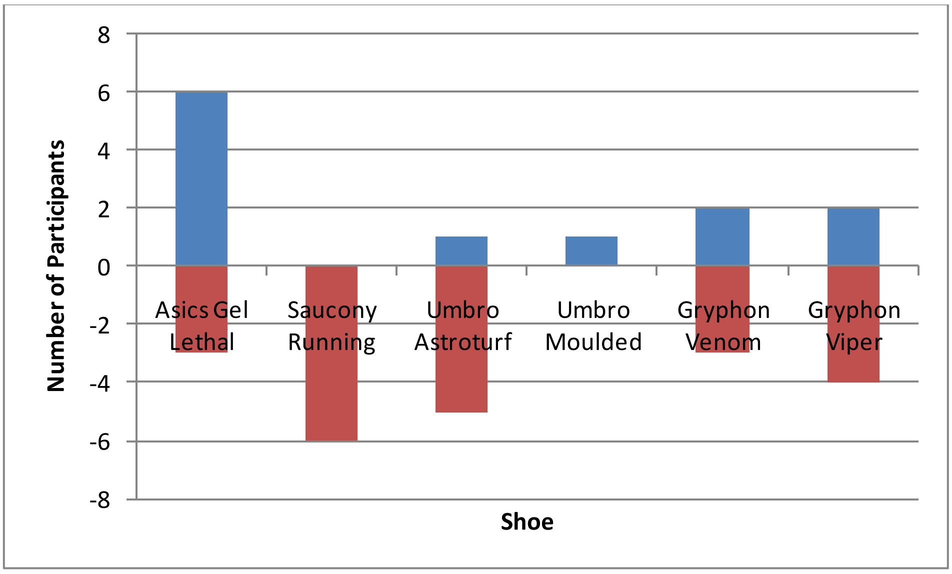
b) Plantar region of the 1st metatarsal head during running

Figure 7.5 Number of participants reporting significant differences (P<0.05) in the magnitude of the peak pressure recorded under the 1st metatarsal head, when comparing the effects of each footwear to all other pairs. Blue = lower peak pressure, Red = higher peak pressure[2]

[2] For each participant investigated, each footwear design may have reported a significantly smaller peak pressure compared to another design of footwear, yet reported a larger peak pressure when compared to a further design of footwear. Therefore as there were 8 participants there can be a maximum of 8 higher and 8 lower peak pressures (n=16)



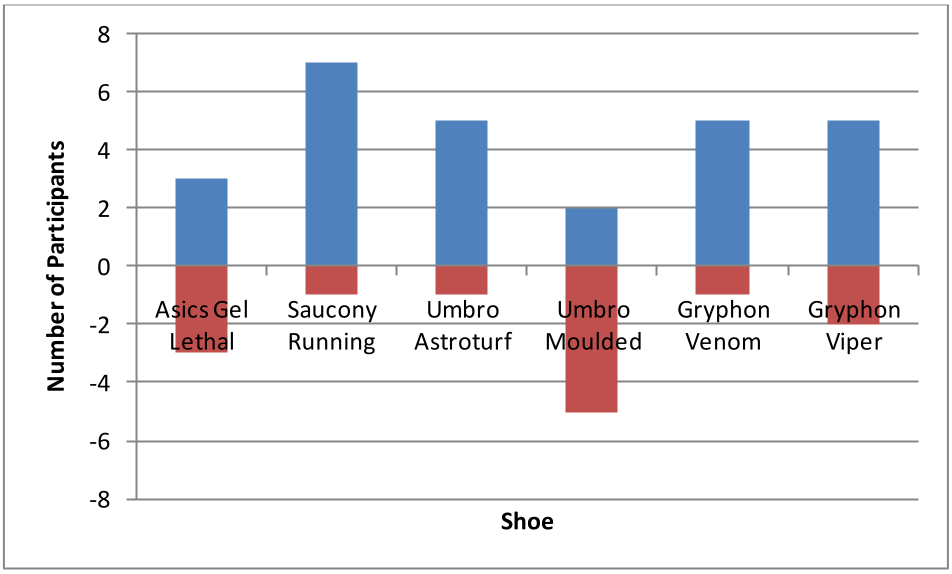
a) **Plantar region of the 2nd and 3rd metatarsal heads during jogging**



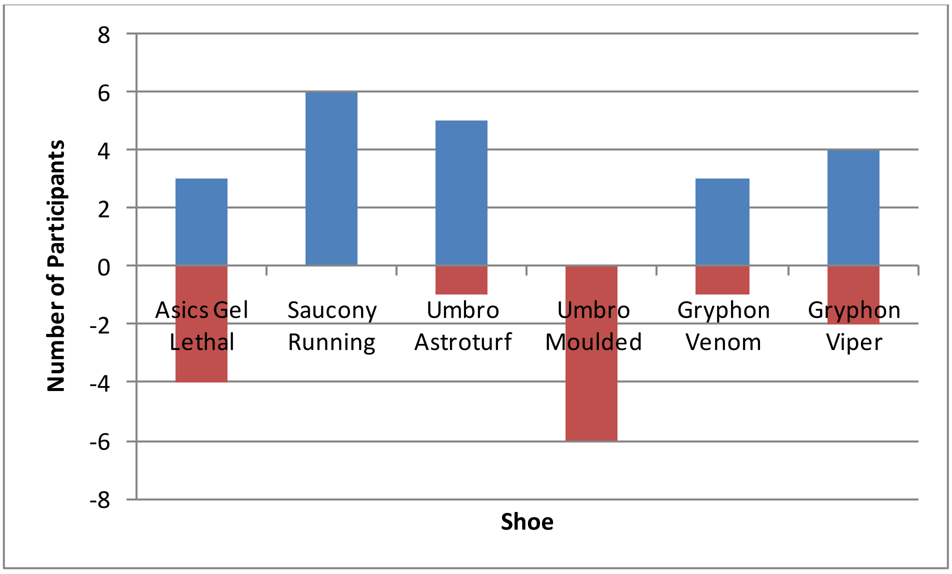
b) **Plantar region of the 2nd and 3rd metatarsal heads during running**

Figure 7.6 Number of participants reporting significant differences ($P<0.05$) in the magnitude of the peak pressure recorded under the 2nd and 3rd metatarsal heads, when comparing the effects of each footwear to all other pairs. Blue = lower peak pressure, Red = higher peak pressure [3]

[3] For each participant investigated, each footwear design may have reported a significantly smaller peak pressure compared to another design of footwear, yet reported a larger peak pressure when compared to a further design of footwear. Therefore as there were 8 participants there can be a maximum of 8 higher and 8 lower peak pressures (n=16)



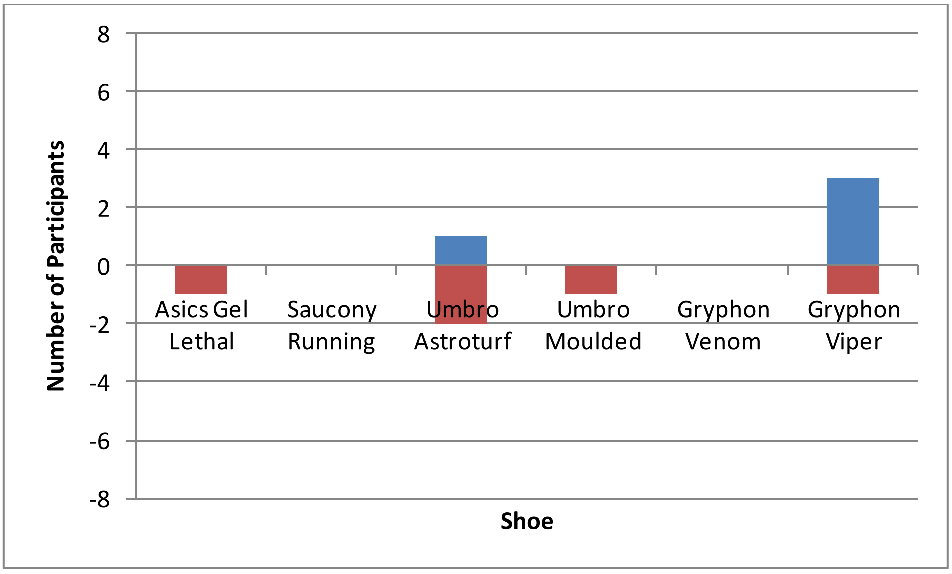
a) Plantar region of the 4th and 5th metatarsal heads during jogging



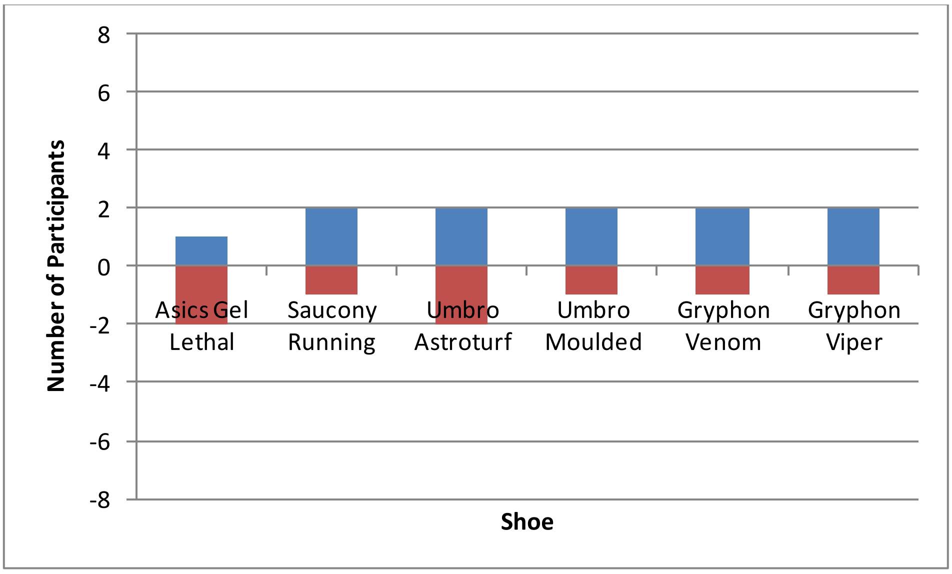
b) Plantar region of the 4th and 5th metatarsal heads during running

Figure 7.7 Number of participants reporting significant differences (P<0.05) in the magnitude of the peak pressure recorded under the 4th and 5th metatarsal heads, when comparing the effects of each footwear to all other pairs. Blue = lower peak pressure, Red = higher peak pressure [4]

[4] For each participant investigated, each footwear design may have reported a significantly smaller peak pressure compared to another design of footwear, yet reported a larger peak pressure when compared to a further design of footwear. Therefore as there were 8 participants there can be a maximum of 8 higher and 8 lower peak pressures (n=16)



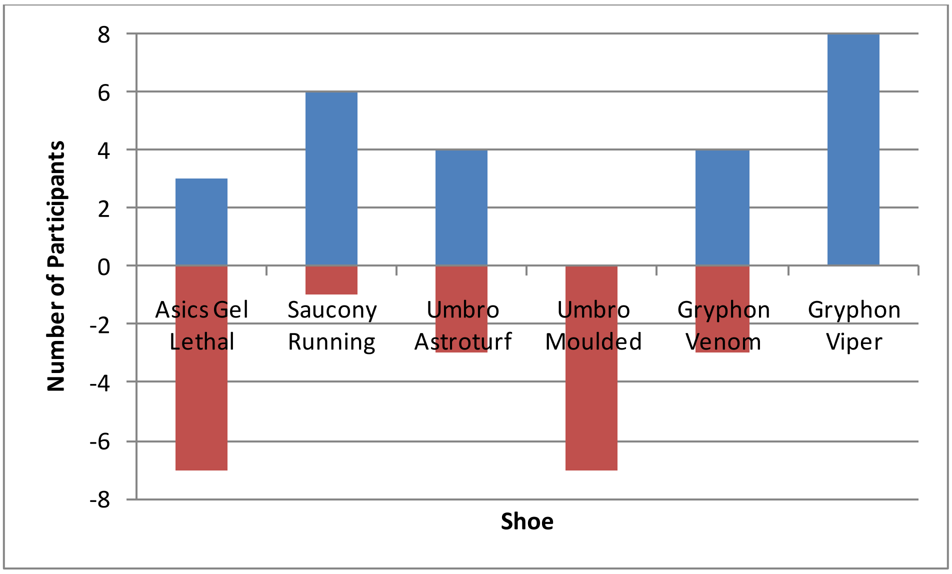
a) Plantar region of the calcaneum during jogging



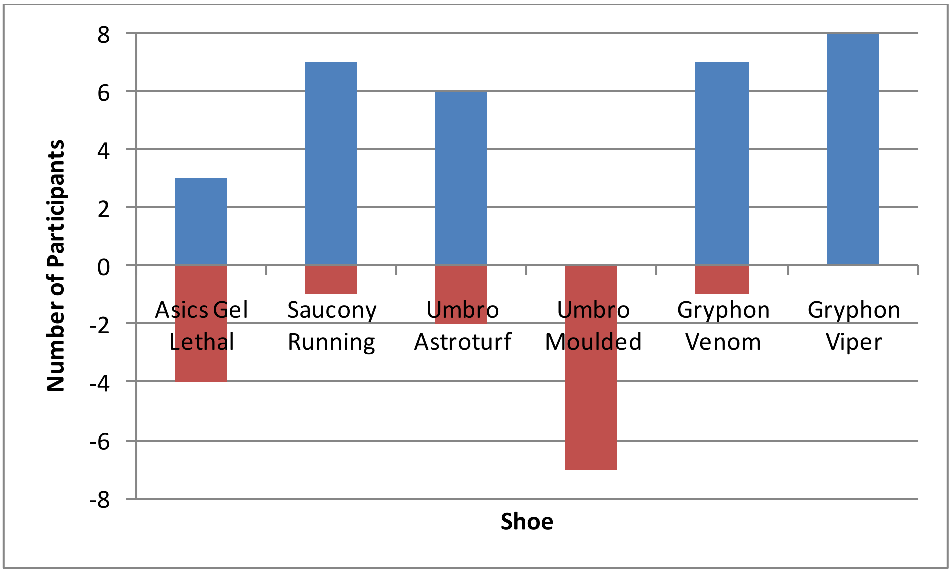
b) Plantar region of the calcaneum during running

Figure 7.8 Number of participants reporting significant differences (P<0.05) in the magnitude of the peak pressure recorded under the calcaneum, when comparing the effects of each footwear to all other pairs. Blue = lower peak pressure, Red = higher peak pressure[5]

[5] For each participant investigated, each footwear design may have reported a significantly smaller peak pressure compared to another design of footwear, yet reported a larger peak pressure when compared to a further design of footwear. Therefore as there were 8 participants there can be a maximum of 8 higher and 8 lower peak pressures (n=16)



a) Entire plantar region of the foot during jogging



b) Entire plantar region of the foot during running

Figure 7.9 Number of participants reporting significant differences ($P<0.05$) in the magnitude of the peak pressure recorded under the entire plantar region, when comparing the effects of each footwear to all other pairs. Blue = lower peak pressure, Red = higher peak pressure[6]

[6] For each participant investigated, each footwear design may have reported a significantly smaller peak pressure compared to another design of footwear, yet reported a larger peak pressure when compared to a further design of footwear. Therefore as there were 8 participants there can be a maximum of 8 higher and 8 lower peak pressures (n=16)

The deep cleats in the design of the Asics Gel Lethal also produces significantly higher peak pressures across most of the conditions although similarly to the Umbro Moulded reported favourable results at the plantar region of the 2nd and 3rd MTHs while running in the footwear. However in the data recorded during jogging, 6 out of the 8 participants experienced higher peak pressure to at least one of the other shod conditions.

Generally the four other shoes performed well across the areas of the specific plantar regions. This is demonstrated by comparing within participant entire plantar region peak pressure data, with the Saucony Running, Umbro Astroturf, Gryphon Venom, and Gryphon Viper all producing more significantly ($P<0.05$) smaller peak pressure values compared to larger values for more participants at each locomotion speed.

The results for the peak plantar pressure under the calcaneum (Figure 7.8) reported few significant results within participant data between the shoes. This suggests that while the footwear being investigated can be recommended in terms of likelihood to cause a reduction in plantar peak pressures in the forefoot, the calcaneum appears to be less affected by the change in footwear conditions for individuals.

Table 7.2 introduces a further method of analysing the differences between the peak pressure results reported in this study. Each shoe is directly compared to each of the other shoes individually. If a significant difference within a participant's data is present, it is reported as either a larger or a smaller peak pressure. Where there are smaller significant peak pressures reported for the shoe being evaluated the data is coloured red. Where larger detrimental peak pressures are reported, the data is coloured yellow. Therefore in general, for the shoe being investigated, the more red data for a condition suggests the influence of the footwear has been positive in terms of reducing peak pressures for more participants compared to the other shoe. This method of presenting the results allows direct comparison between shoes. Although it is clear that the footwear influences individuals differently, this data provides evidence that could assist in choosing footwear that is more likely to have a positive effect on loading of the plantar surface.

The results show that the Saucony Running (Table 7.2b) presented favourable results in most areas compared to all the other footwear. This was followed by the Gryphon Viper (Table 7.2f) where results were favourable compared to the remaining shoes. For the hockey specific footwear the Asics Gel Lethal (Table 7.2a) clearly reported the least favourable results, with the Umbro Moulded shoe (Table 7.2d) producing the least favourable out of all the footwear. These results could be used to assist in making suitable footwear choices for field hockey participation. It is clear from these results that when considering choosing from the footwear investigated, an individual should be assessed while wearing the shoes. This is because even the shoes identified as being the most favourable generally, for some participants reported a significant increase in peak pressure at various sites of the plantar surface.

The differences seen between running and jogging were varied with the 2nd and 3rd MTHs being the only area where there was an increase in the peak pressure when running in each shoe. The peaks over the entire plantar region suggested that the Gryphon Venom and Gryphon Viper shoes exposed the participants to larger increases in already relatively high values as the velocity increased. These findings agree with previous results that have found an increase in locomotion speed had different effects on the specific areas under the foot (Segal *et al.*, 2004;Taylor *et al.*, 2004). In terms of footwear choice, it would appear that an athlete wanting to protect a certain part of the plantar region could select footwear that would do this. However a consideration towards the breakdown of their locomotion speeds within their sport should be considered.

While previous studies investigating sports specific movements such as a lunge in fencing have found significant ($P<0.05$) reductions in peak pressure across a population (Geil, 2002), such movements are inherently different to forward locomotion. During human locomotion over a uniform flat surface, the body is aware of the forces being applied to the musculoskeletal system in each step and can therefore make the relevant adjustments as recorded during locomotion where a reduction in cushioning is available (Aguinaldo and Mahar, 2003;De Wit *et al.*, 2000;Hennig *et al.*, 1996). Therefore across a general population, detrimental kinetic data may demonstrate non significant values due to the individuals adopting various different movement strategies. However significant differences at the $P<0.05$ level, have been reported in peak plantar pressures during locomotion when comparing the affects of footwear (Wiegerinck *et al.*, 2009;Hennig and Milani, 1995). Although this research was carries out in accordance to previously published guidelines for trial sizes (Bates *et al.*, 1992).

Table 7.2 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) peak pressures. Comparing (a) Asics Gel Lethal, (b) Saucony Running, (c) Umbro Astroturf, (d) Umbro Moulded, (e) Gryphon Venom and (d) Gryphon Viper shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

a) Asics Gel Lethal

		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper	
Value of the Asics Gel Lethal Variable Below, Compared to the Condition Above		S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	2	2	1	6	0	1	0	0	2
	Running on ISSS	0	2	1	2	4	2	2	2	0	2
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	0	5	0	2	0	4	0	1	0	6
	Running on ISSS	0	4	1	2	1	0	1	1	0	3
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	0	2	0	1	3	1	1	1	1	2
	Running on ISSS	0	2	0	1	2	0	0	0	1	2
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	2
	Running on ISSS	0	3	0	1	4	0	0	1	0	3
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	4	1	0	3	0	0	0	0	5
	Running on ISSS	0	1	0	1	0	2	0	1	0	2

b) Saucony Running Shoe

		Asics Gel Lethal		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper	
Value of the Saucony Running Variable Below Compared to the Condition Above:		S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	2	0	2	0	7	0	1	0	0	0
	Running on ISSS	2	0	2	1	7	0	1	1	1	0
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	5	0	2	0	2	0	3	0	0	1
	Running on ISSS	4	0	1	0	2	0	2	0	1	0
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	2	0	1	0	5	1	1	0	2	0
	Running on ISSS	2	0	1	0	5	0	1	0	2	0
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0
	Running on ISSS	1	0	0	0	0	1	0	0	0	1
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	4	0	2	0	5	0	1	0	0	1
	Running on ISSS	3	0	2	0	6	0	1	0	0	1

c) Umbro Astroturf Shoe

Value of the Umbro Astroturf Variable Below, Compared to the Condition above :		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L	Gryphon Viper S	Gryphon Viper L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	1	2	0	2	6	0	1	2	0	1
	Running on ISSS	2	1	2	1	6	0	1	0	0	1
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	2	0	0	2	2	1	2	0	0	2
	Running on ISSS	2	1	0	1	1	0	1	0	0	1
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	0	0	1	5	0	0	1	1	1
	Running on ISSS	1	0	0	1	4	0	1	0	2	0
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	1	0	0	1	2
	Running on ISSS	1	0	0	0	0	1	1	2	0	1
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	1	0	2	4	0	0	1	0	3
	Running on ISSS	1	0	0	2	6	0	1	1	0	1

d) Umbro Moulded Shoe

Value of the Umbro Moulded Variable below, Compared to the Condition above :		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Gryphon Venom S	Gryphon Venom L	Gryphon Viper S	Gryphon Viper L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	6	0	7	0	6	1	4	0	7
	Running on ISSS	0	6	0	7	0	6	0	6	0	6
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	4	0	0	2	1	2	1	1	0	3
	Running on ISSS	0	1	0	2	0	1	0	2	0	2
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	3	1	5	0	5	0	4	0	5
	Running on ISSS	0	2	0	5	0	4	0	2	0	4
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	1	0	0	0	0	0
	Running on ISSS	2	0	1	0	1	0	1	0	1	1
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	3	0	5	0	4	0	4	0	7
	Running on ISSS	0	4	0	6	0	6	0	4	0	7

e) Gryphon Venom Shoe

Value of the Gryphon Venom Variable Below, Compared to the Condition above :		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Viper S	Gryphon Viper L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	1	0	1	1	2	4	1	0	3
	Running on ISSS	2	2	1	1	0	1	6	0	0	0
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	1	0	0	3	0	2	1	1	0	3
	Running on ISSS	1	1	2	0	1	0	2	0	0	1
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	1	0	1	1	0	4	0	0	0
	Running on ISSS	0	0	0	1	0	1	2	0	1	0
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0
	Running on ISSS	0	0	0	0	2	1	0	1	0	0
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	0	0	1	1	0	4	0	0	2
	Running on ISSS	1	0	0	1	1	1	4	0	0	1

f) Gryphon Viper Shoe

Value of the Gryphon Venom Variable Below, Compared to the Condition above :		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	2	0	0	0	1	0	7	0	3	0
	Running on ISSS	2	0	0	1	1	0	6	0	0	0
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	6	0	1	0	2	0	3	0	3	0
	Running on ISSS	3	0	0	1	1	0	2	0	1	0
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	2	1	0	2	1	1	5	0	0	0
	Running on ISSS	2	1	0	2	0	2	4	0	0	1
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	1	0	0	0	0	0
	Running on ISSS	2	0	1	0	1	0	1	1	0	0
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	5	0	1	0	3	0	7	0	3	0
	Running on ISSS	3	0	1	0	1	0	7	0	1	0

Due to the individual variability between participants and the calibration issues with the technology, a larger population of participants may produce more conclusive results.

7.4 Conclusion

Due to the considerable intra-participant variability, this method of measuring in-shoe pressure may have limited application in terms of designing a shoe to reduce levels of localised pressure. However for an individual, results of this study show that footwear used in field hockey participation can affect the distribution of localised pressure applied to the plantar region during sports specific movements.

The effects of footwear on the pressure applied to the plantar appear to be influenced by the individual; an individual assessment therefore, may be required for each participant. However the evidence from this study does demonstrate there are favoured footwear designs that can inform footwear choice. It is also important to highlight that other issues such as GRFs and impact shock need to also be considered when selecting hockey footwear that may reduce overuse injury prevalence. Footwear causing a reduction in an injury causing factor in the plantar surface of the foot, may increase the magnitude of another injury causing factor Therefore any footwear choices made need to be assessed for the other factors that are linked to overuse injuries such as GRF characteristics and impact acceleration measurements.

Chapter Eight

Effects of Footwear on Ground Reaction Forces

8 Effects of footwear on ground reaction forces

This study will aim to identify if any of the footwear being investigated in this thesis, has an effect on the overuse injury influencing GRFs, measured using an in-shoe sensor during locomotion.

8.1 Introduction

During locomotion in field hockey participation the musculoskeletal system is subjected to loading from GRFs which if excessive can expose participants to an increase in the likelihood of sustaining an injury and therefore reduce participation (Hamill *et al.*, 1983;Kaplan and Heegaard, 2000;McClay *et al.*, 1994;Clarke *et al.*, 1983a;Frederick and Hagy, 1986;Munro *et al.*, 1987;Nilsson and Thorstensson, 1989;Collins and Whittle, 1989;Wiegerinck *et al.*, 2009;Keller *et al.*, 1996;Lees and Field, 1985). Peak GRFs occurring shortly after foot to ground contact often produce a noticeable impact peak (Figure 5.3). During this period, the musculoskeletal system is exposed to a transient impact shock has been linked to overuse injuries and can be measured using accelerometers mounted to the body (Auvinet *et al.*, 2002a;Lafortune and Hennig, 1992;Voloshin *et al.*, 1998;Verbitsky *et al.*, 1998;Lafortune and Hennig, 1991;Lafortune *et al.*, 1995a;Hennig *et al.*, 1993;Mercer *et al.*, 2002;Hreljac, 2004;Zhang *et al.*, 2008;Auvinet *et al.*, 2002b).

Various GRF variables have been reported in previous research (Chapter 5). Loading rate characteristics have been highlighted in previous research as being strongly correlated to tibial accelerations (Hennig and Lafortune, 1991;Laughton *et al.*, 2003).

The instantaneous loading rate was reported to have the strongest correlation (r=0.469, P<0.001) to the peak tibial acceleration, in those reported in Chapter 5. Laughton *et al.* (2003) reported similar results for the average loading rate in heel strike running (r=0.47, P<0.05) and found a stronger correlation in peak instantaneous loading rate (r=0.70, P<0.05). It would appear from previous published research and the data reported in Chapter 5 of this thesis that the loading rates provide the best evidence of GRFs exposing the musculoskeletal system to overuse injuries. However, the results from Chapter 5 identified the strongest overall correlation (r=-0.526, P<0.001) between tibial accelerations and the time to reach the second peak typically seen in vertical force data. Therefore this variable will be considered important in this study.

Most of the research investigating GRFs discussed in this thesis has used force plates to collect accurate kinetics (Dayakidis and Boudolos, 2006;Hennig and Lafortune, 1991;Stuelcken and Sinclair, 2009;Gottschall and Kram, 2005;Nachbauer and Nigg, 1992;Nilsson and Thorstensson, 1989). When comparing force data collected with an F-scan® in-shoe system to force plate data, high correlations (r = 0.93) between the two sets of data were reported (Mueller and Strube, 1996). The high correlations indicated that the output of the F-scan system changes in linear fashion to force plate values. However an offset that produced lower magnitudes throughout recordings was reported as present. Results from research comparing force plate and in-shoe pressure systems have concluded that the sum of the in-shoe pressure can give a good estimation of the vertical GRF recorded from a force plate during human locomotion (Hennig and Milani, 1995). Although there is a relatively small inherent inaccuracy in measuring force data in this manner, by using the same calibration between shoes for each participant and careful in-shoe fitting, this methodology allows comparisons of the estimated vertical GRF through the footwear being investigated. A major advantage of using this system to estimate force is that the detrimental effects on data caused by force plate targeting (Challis, 2001) are eliminated. Furthermore by using such a system, multiple foot to ground impacts can be recorded during each run performed by a participant (Forner Cordero *et al.*, 2004). Collecting data in this manner compared to the use of a single force plate requires less runs to be performed by participants. For this investigation these factors were important as the participants were being asked to jog ($3.3 m.s^{-1}$) and run ($5.0 m.s^{-1}$) in six different types of footwear until at least nine data sets were recorded. Using an in-shoe device recording three data sets per run, meant that each speed and footwear condition could be suitably collected in three good runs. This meant that the minimum amount of runs a participant had to perform was relatively small (n=36), when compared to the number of minimum runs that collection of the same amount of data would require using a single force plate (n=108). Due to the issues with force plate targeting, this number could be much higher often depending on the participant's ability to consistently strike the force plate with the entire sole of the shoe. Being efficient can be important when using shared laboratories and also when attempting to recruit participants. Furthermore, the in-shoe device methodology reduces the effects of fatigue which has been shown to influence kinematic and kinetic data recorded (Mizrahi *et al.*, 2000a;Tsai *et al.*, 2009;Mizrahi *et al.*, 2000c;Coventry *et al.*, 2006;Bisiaux and Moretto, 2008;Schlee *et al.*, 2006;Nummela *et al.*, 1996;Derrick *et al.*, 2002). The reduction in the onset of fatigue in participants should assist in collecting more valid data between shod conditions. However using the in-shoe pressure system has been shown to alter gait characteristics during running on a treadmill with significant ($P<0.001$) decreases in stride length and increases in stride frequency reported when wearing the system (Kong and De Heer, 2009). This would appear to suggest a stability issue. By comparing shoes all with inserts in, a fair comparison can be reported, although a system that has no significant influence on human locomotion should be the aim for future technologies.

The aim of this study was to investigate the effects that various footwear typically worn by field hockey participants, has on the characteristics of the GRF applied to the plantar surface of the foot during the ground contact phase of locomotion.

8.2 Methodology

The data for this study was collected under the same methodology with the same group of participants as in section 7.2.

8.2.1 Data processing and statistics

Using the Tekscan software (Tekscan inc, MA, USA), the entire loading of the foot plantar surface was exported for the duration of each of the selected stance phases. Due to the high correlation ($r=0.93$) between the linear characteristics of the pressure data using an F-Scan® system (Tekscan inc, MA, USA), and the vertical GRF characteristics recorded using force plates (Mueller and Strube, 1996), the outputs were considered a reasonable estimate of the vertical GRF. The vertical GRF estimations were analysed and processed through a Matlab software program (adapted from the one developed in Chapter 6) that required the user to identify or best estimate the 1st force peak. The program calculated various peak forces, loading rates and time events. ANOVAs were used to compare the affects of the different footwear being tested, comparing the mean values reported across the population of participants as well as the differences reported for within-participant data.

8.3 Results and discussion

The results from this study demonstrated similar typical GRF characteristics (Figures 8.1 and 8.2) when compared to those obtained in Chapter 5 of this thesis (Figure 5.3 and 5.4). However the mean values presented across the GRF peaks and loading rates were considerably smaller (Table 8.1) compared to the results from Chapter 5 (Table 5.1) and those from previous research (Nilsson and Thorstensson, 1989;McClay *et al.*, 1994;Cavanagh and Lafortune, 1980). The times to peak loading rate and 1st impact peak were approximately the same as those reported in the earlier study. A small part of the differences in the magnitude of the loading rates and the peaks may be due to the lower frequency at which the data in this study was recorded (500Hz) compared to the earlier study (1000Hz). By sampling at a lower frequency, some high frequency components that may indicate transient forces could be missing from the data (Aguinaldo and Mahar, 2003). Furthermore, the peak loading rate is calculated between data points so at 500Hz it is recorded over 2ms compared to 1ms at 1000Hz. This would effectively report an average loading rate over two 1ms recordings, so the peak value over 1ms would be reduced to the average of the two 1ms periods. However, the offset of data, reported in previous literature (Mueller and Strube, 1996) was clearly apparent and the main cause of the lower magnitude of force from that recorded from force plate systems. This means that the actual magnitudes of values involving GRF values such as impact peaks and loading rates cannot be compared between participants. However as this study is investigating differences between footwear and for each participant the same insole and calibration is used. Therefore the relative differences between GRF characteristics can be used to identify footwear that would produce more favourable kinetics in terms of injury prevention.

Running in all shoes compared to jogging reported significantly ($P<0.05$) larger values for all of the analysed methods of recording loading rates (Table 8.1). The magnitude of the first vertical force peak was significantly ($P<0.05$) larger for the running conditions and the peak was also reached in a significantly ($P<0.05$) shorter mean time. These results agree with previous literature that highlighted increases in speed as placing the musculoskeletal system under increased detrimental loading (Perry and Lafortune, 1995;Burnfield *et al.*, 2004). This larger force peak that was reached in a shorter time, producing higher rates of loading suggests that at higher speeds, the body is more exposed to overuse injuries as previously reported.

The comparisons between the speed of locomotion and the shoes for the various kinetic variables are shown in Figures 8.3 to 8.5. The variability in the data is shown across the box plots, which show the median values, the 50% data and the 100% data range (excluding outliers).

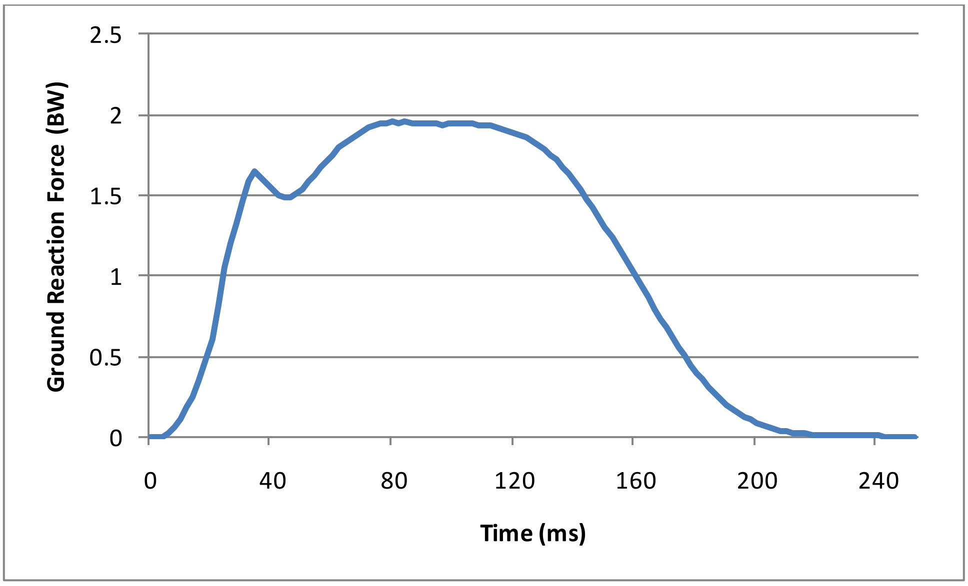


Figure 8.1 Typical vertical force data during stance

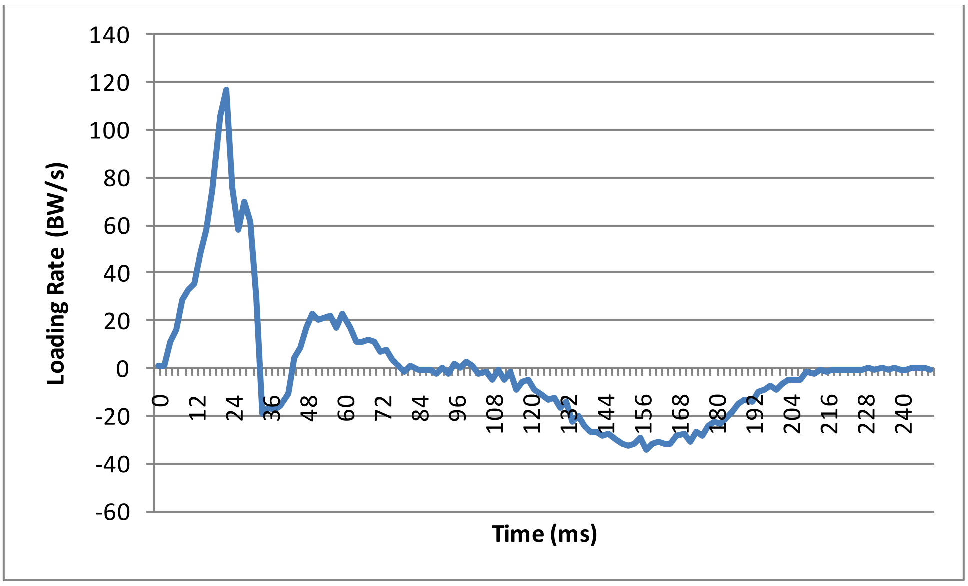


Figure 8.2 Typical vertical loading rate data during stance

Table 8.1 Mean Peak Ground Reaction Force Characteristics for all Participants

Condition		TPVLR (ms)		TVFP1 (ms)		TVFP2 (ms)		VFP1 (BW)		VFP2 (BW)		AVLR ($BW.s^{-1}$)		AVL20T90 ($BW.s^{-1}$)		AVL20T80 ($BW.s^{-1}$)		AVL50NT50 NBW ($BW.s^{-1}$)		PVLR ($BW.s^{-1}$)	
Asics Gel Lethal	Jogging	16.0	±7.6	32.5	±9.3	117.5	±19.2	1.2	±0.4	2.2	±0.2	36.4	±6.9	45.1	±13.1	48.7	±14.2	42.9	±10.3	63.9	±13.7
	Running	16.3	±3.1	28.9	±8.0	86.0	±13.1	1.4	±0.2	2.3	±0.4	54.3	±20.9	72.4	±35.7	76.3	±37.4	64.9	±24.2	99.4	±37.6
Saucony Running	Jogging	18.8	±4.3	37.2	±9.0	108.5	±25.3	1.5	±0.4	2.3	±0.3	43.6	±16.3	64.9	±41.4	67.5	±43.8	51.7	±29.2	81.0	±48.9
	Running	23.3	±11.2	32.3	±8.3	92.5	±18.1	1.5	±0.3	2.3	±0.3	47.9	±16.7	65.7	±24.0	69.5	±26.5	53.2	±16.2	84.9	±28.2
Umbro Astroturf	Jogging	21.5	±9.0	32.8	±9.9	111.3	±23.2	1.3	±0.4	2.4	±0.4	42.4	±17.8	60.2	±42.9	61.7	±42.9	49.6	±32.0	82.5	±56.4
	Running	21.4	±7.8	32.6	±9.4	88.8	±15.8	1.6	±0.4	2.4	±0.4	51.9	±12.5	68.1	±21.4	70.9	±24.4	59.0	±16.2	99.8	±24.1
Umbro Moulded	Jogging	9.2	±3.4	26.4	±8.3	117.5	±20.2	1.4	±0.5	2.4	±0.6	56.7	±31.0	74.6	±49.0	83.3	±58.6	73.6	±60.5	107.1	±64.0
	Running	8.8	±2.1	22.7	±4.0	88.3	±9.3	1.4	±0.3	2.4	±0.6	69.7	±30.9	102.3	±51.1	113.7	±49.7	99.0	±45.6	144.8	±48.1
Gryphon Venom	Jogging	22.8	±7.5	37.2	±10.6	121.9	±22.4	1.4	±0.3	2.3	±0.5	40.5	±19.1	55.0	±33.3	58.0	±35.2	43.8	±18.4	73.8	±41.0
	Running	16.7	±2.6	30.9	±6.6	88.6	±10.3	1.6	±0.4	2.3	±0.4	57.1	±27.2	81.8	±53.1	87.0	±55.4	65.3	±22.1	108.1	±57.7
Gryphon Viper	Jogging	26.0	±6.2	35.1	±8.8	118.8	±21.6	1.3	±0.4	2.2	±0.3	38.0	±13.3	49.7	±22.9	52.3	±25.3	40.1	±17.1	70.1	±28.7
	Running	24.8	±8.9	31.2	±8.5	87.9	±13.1	1.5	±0.4	2.3	±0.4	50.3	±19.1	68.6	±32.5	71.8	±35.5	56.5	±23.2	105.9	±39.0

Comparing the influence of the footwear during running for the population of participants, significant ($P<0.05$) lower PVLR and AVL50NT50NBW were reported during running for the Saucony Running Shoe (84.9 ±28.2BW.s^{-1} and 53.2 ±16.2BW.s^{-1} respectively) compared to the Umbro Moulded shoe (144.8. ±48.1 BW.s^{-1} and 99.0 ±45.6 BW.s^{-1} respectively). Furthermore, the Umbro Astroturf shoe (59.0 ±16.2 BW.s^{-1}) and Gryphon Viper (56.5 ±23.2 BW.s^{-1}) reported significantly ($P<0.05$) lower values for the AVL50NT50NBW compared to the Umbro Moulded shoes. The only other variable reporting a significant difference was the TPVLR. Figure 8.4 shows that the distribution of the Umbro Moulded shoe values demonstrates a consistently shorter time to the peak loading rate than the majority of the other footwear conditions. During running, this timing variable reported a significant ($P<0.05$) decrease in time for the Umbro Moulded condition compared to the Saucony Running (23.3 ±11.2 s), Umbro Astroturf (21.4 ±7.8 s), and the Gryphon Viper (24.8 ±8.9 s). Significant differences ($P<0.05$) were also reported for the TPVLR variable during jogging. Again the Umbro Moulded condition reported a shorter period of time to the peak from foot down (9.2 ±3.4 s) this time compared to the Umbro Astroturf (21.5 ±9.0 s), Gryphon Venom (22.8 ±7.5 s), and Gryphon Viper (26.0 ±6.2 s). Although the TPVLR is not identified in the previous literature discussed, within this thesis, from the timing variables investigated (Table 5.2), TPVLR reported the strongest correlation ($r=-0.359$, $P<0.001$) with the impact accelerations measured at the tibia. When compared to the TVFP1 ($r=-0.336$, $P<0.001$) it is a stronger correlation using the methodology in this thesis. In previous research the TVFP1 has been found to be strongly correlated ($r=-0.89$) to the peak tibial acceleration (Hennig and Lafortune, 1991). Therefore the TPVLR variable may be worth considering in future research. No other significant values between the shoes were reported. The distribution of the data shown in Figures 8.3-8.6 shows the spread of the mean values suggests large variability between the influence of footwear on individual's GRF characteristics. This agrees with conclusion drawn from previous study that suggest footwear effects on kinetics are dependent mainly on the individual and in general cannot be applied to a general population (Kersting and Bruggemann, 2006). The three variables (TPVLR, PVLR and AVL50NT50NBW) that reported significant differences across the population when comparing footwear conditions do not require the identification of a force peak to be calculated. These results provide evidence that methodologies such as identifying the loading rate over an increase in force of a set value such as a BW or between frames is a more suitable GRF variable to report.

The results of the mean data across all participants from this study compared to the mechanical drop test (Chapter 6) show that as in the mechanical drop test (Figures 6.12 and 6.13), and human locomotion test (Figures 8.5 and 8.6), the largest peak rates of loading were identified in the Umbro Moulded footwear. In the mechanical test the Umbro Moulded footwear clearly provides much less cushioning than the other shoes. In the human testing the distinction is not so clear. The other footwear produced many significant differences during the drop test but none in the human running tests. Previous research investigating the effects of midsole hardness on loading, found significant differences were only found in extreme alterations of midsole hardness (Kersting and Bruggemann, 2006). This would agree with the results from this study and suggests in shod conditions the human body has the ability to attenuate excessive loading. From the drop test results the mean PVLR value of the Umbro Moulded footwear is over 4 times the amount of the Gryphon

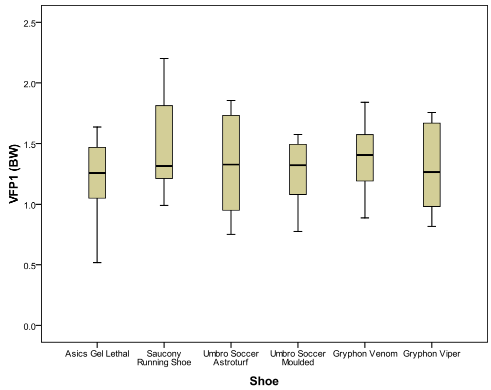
Viper, Gryphon Venom, Asics Gel Lethal and Saucony Running shoes. However, during the human testing it was only 30-70% larger. This demonstrates the human body's ability to attenuate excessive loading. Further evidence of this is demonstrated by the Umbro Astroturf footwear. Comparing the PVLR from the mechanical tests to the same four footwear, the Umbro Astroturfs mean PVLR is over twice the magnitude in the mechanical test yet in the human test has a lower magnitude than the Gryphon Viper and Gryphon Venom, an almost identical value as the Asics Gel Lethal and is 18% larger than the Saucony Running shoes. It is worth pointing out that the lowest PVLR is in the Saucony Running shoe which matches the same order as in the mechanical test and the results from Kersting and Bruggemann (2006). It would appear that only in extreme midsole hardness can the body not adapt its movement strategy enough and even then large differences reported in mechanical tests are reduced in human locomotion testing. It appears as would agree with previous research, that the effects of footwear cushioning are very dependent on the individual participant (Kersting and Bruggemann, 2006).

Figures 8.7 to 8.14 show the number of participants' data that reported a significant difference between the footwear condition and any of the other five footwear choices. The influence of footwear for each participant was found to significantly affect many of the GRF variables reported (Figures 8.7 to 8.14). While the TVFP1 only reported significant differences for a maximum of 2 participants for any footwear condition, the TVFP2 and the TPVL variables demonstrates that the Gryphon Venom and Gryphon Viper would be the more favourable choice with the Umbro Moulded the least favourable when considering these variables' relationship to tibial accelerations. This pattern generally follows amongst the various loading rate variables (Figures 8.10 to 8.14), with favourable results also generally reported for all shoes except the Umbro Moulded Shoes. The positive results for the shoes were in nearly all cases when compared to the Umbro Moulded shoes. This gives further evidence that these shoes place participants at increased risk of overuse injury.

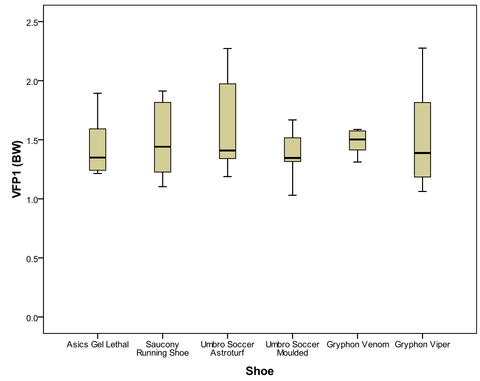
Another factor that should be considered is that more significant differences were found in the jogging conditions compared to the running conditions. This is a important finding in terms of field hockey participation, as a greater amount of time is spent jogging compared to running (40.5% and 5.6% respectively) (Spencer et al., 2004b). If during jogging, impact loading is still relatively high, this will expose participants to much longer periods of being vulnerable to repetitive large magnitude loading, placing the body at increased risk of suffering an overuse injury.

Table 8.2 uses the methodology implemented in the previous chapter (Table 7.2). Each shoe is directly compared to each of the other shoes. If a significant difference within a participant's data is present, it is reported as either a larger or a smaller GRF characteristic. For timing variables a larger value is considered favourable due to smaller values being linked to excessive loading. Therefore the data is coloured red meaning favourable, when more participants report significantly larger timing values, and yellow when more reported smaller values. For loading variables a smaller value is considered favourable and therefore data with smaller values is coloured red and data with larger values is coloured yellow. Therefore in general, for the shoe being investigated, the more red for a condition, the more effective the shoe condition was for individuals than the other shoes. This method of presenting the results allows direct comparison between shoes. Although it is clear that the footwear influences individuals differently, this data provides evidence that could assist in choosing footwear that is more likely to have a positive effect on loading of the plantar surface.

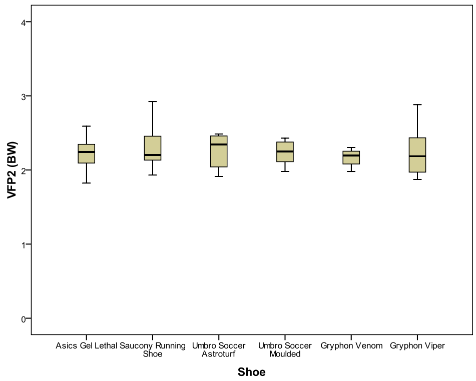
The results are not conclusive however, the influence of the Gryphon Venom and the Gryphon Viper both produce relatively favourable results with the Umbro Moulded producing the least favourable followed by the Umbro Astroturf shoe. The Saucony running shoe that may be expected to be the most effective design for reducing impact loading during locomotion reported unfavourable results compared to the field hockey specific footwear. This evidence suggests that field hockey participants using running shoes which has been seen even at Olympic level (Frederick, 2008), will probably not be gaining further protection from loading of the musculoskeletal system above the foot. However in the previous chapter more favourable results were reported for the Saucony running shoe compared to the field hockey specific footwear. It would appear that footwear may offer reductions in injury potential in one area of the body while exposing another area to an increase risk of injury. For participants with a history of a certain type of injury this information could be very beneficial in making an informed choice.



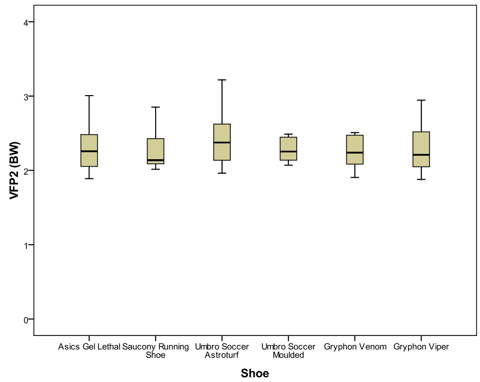
a) Jogging (3.3m.s^{-1})



b) Running (5m.s^{-1})

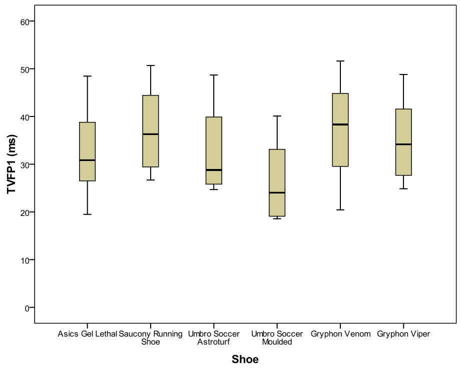


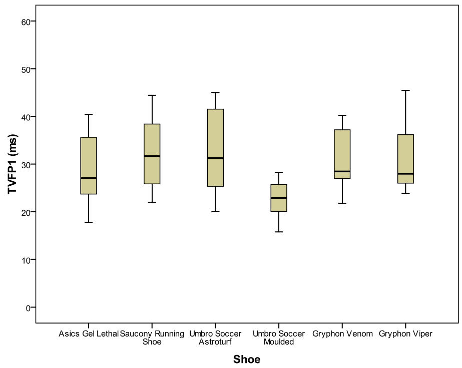
a) Jogging (3.3m.s^{-1})



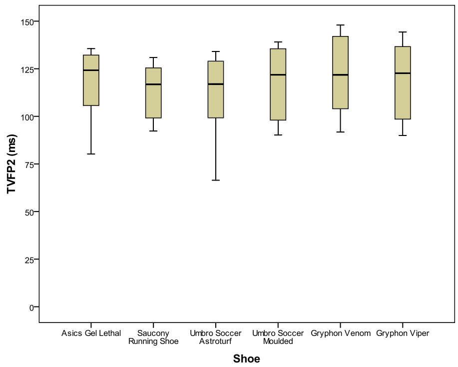
b) Running (5m.s^{-1})

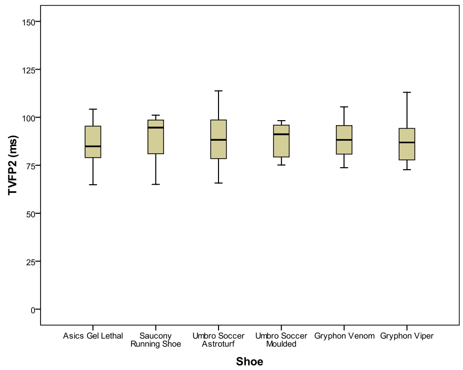
Figure 8.3 Distribution of mean participant values of vertical force peaks during stance for a) Jogging (3.33m.s^{-1}), and b) Running (5ms^{-1})



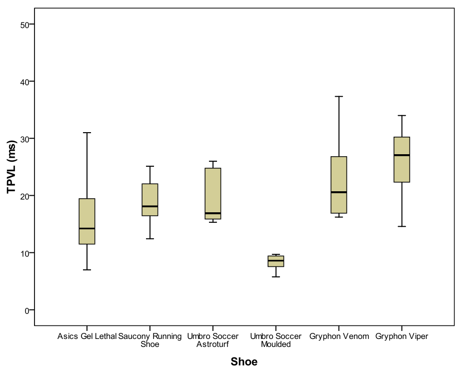


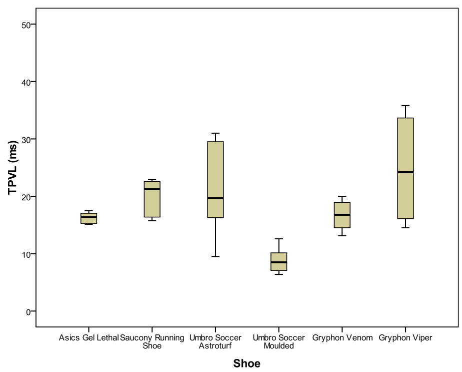
a) Jogging (3.3m.s^{-1}) **b) Running (5m.s^{-1})**





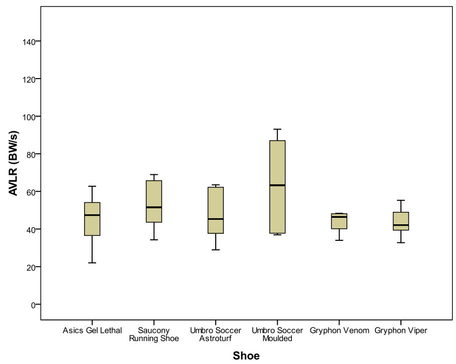
a) Jogging (3.3m.s^{-1}) **b) Running (5m.s^{-1})**



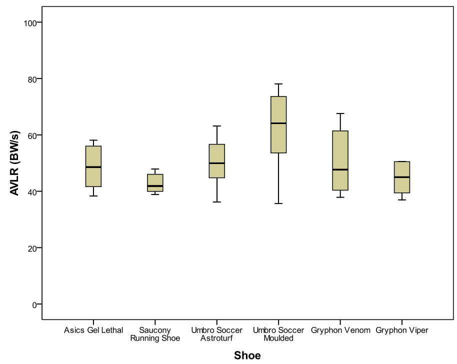


a) Jogging (3.3m.s^{-1}) **b) Running (5m.s^{-1})**

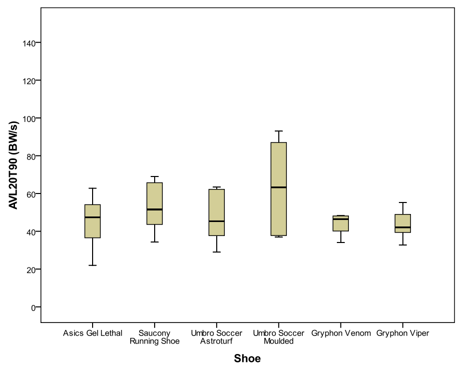
Figure 8.4 Distribution of mean participant values of kinetic event times during stance for a) Jogging (3.33m.s^{-1}), and b) Running (5ms^{-1})



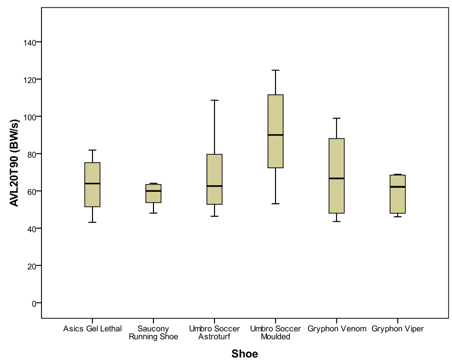
a) Jogging (3.3m.s^{-1})



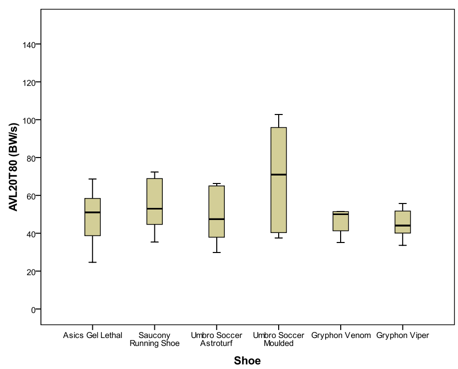
b) Running (5m.s^{-1})



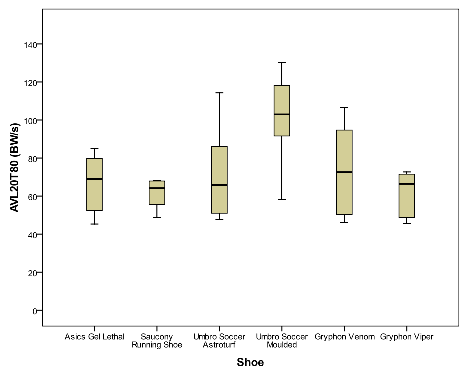
a) Jogging (3.3m.s^{-1})



b) Running (5m.s^{-1})

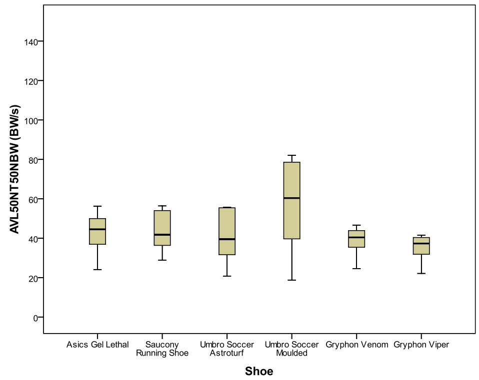


a) Jogging (3.3m.s^{-1})

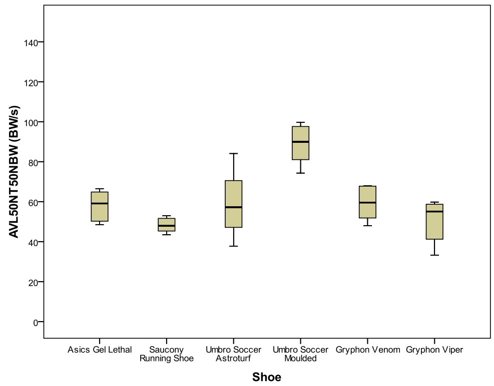


b) Running (5m.s^{-1})

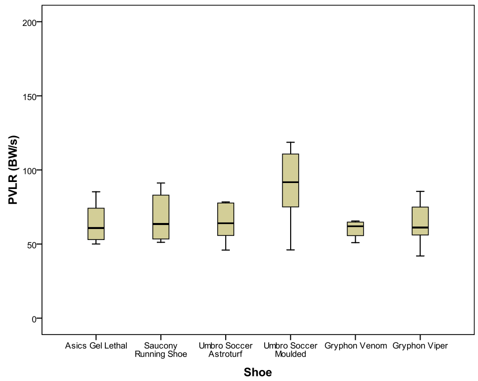
Figure 8.5 Distribution of mean participant values of loading rates during stance for (a) Jogging (3.33m.s^{-1}), and (b) Running (5ms^{-1})



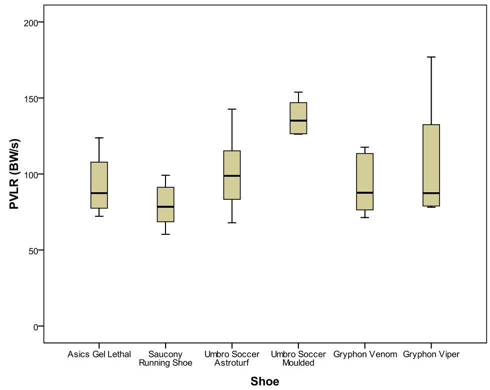
a) Jogging (3.3m.s^{-1})



b) Running (5m.s^{-1})

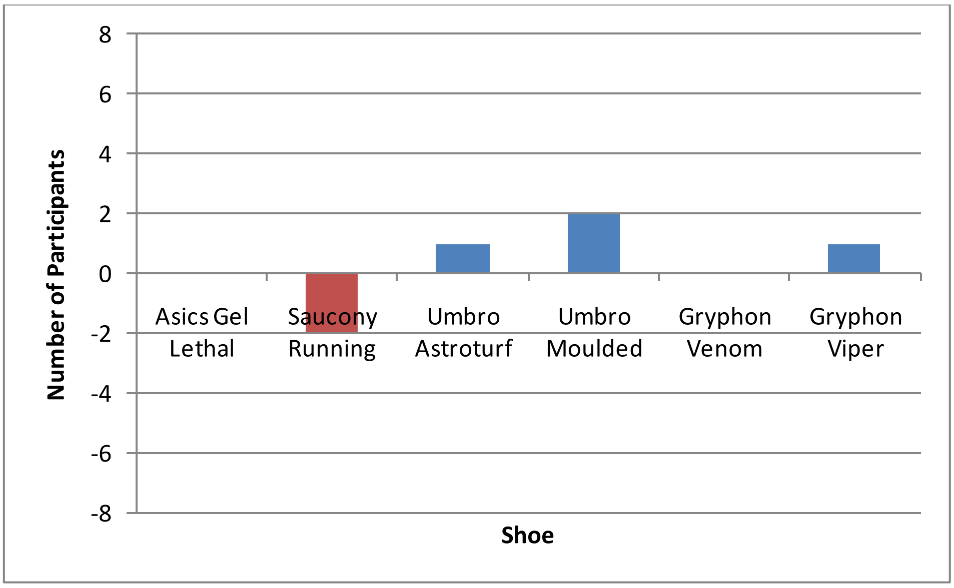


a) Jogging (3.3m.s^{-1})

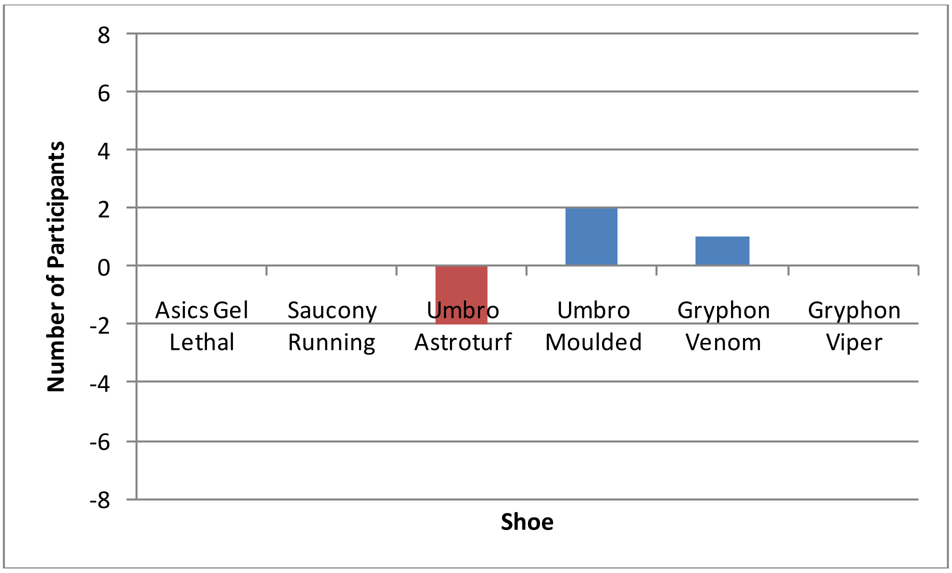


b) Running (5m.s^{-1})

Figure 8.6Distribution of mean participant values of loading rates during stance for (a) Jogging (3.33m.s^{-1}), and (b) Running (5ms^{-1})



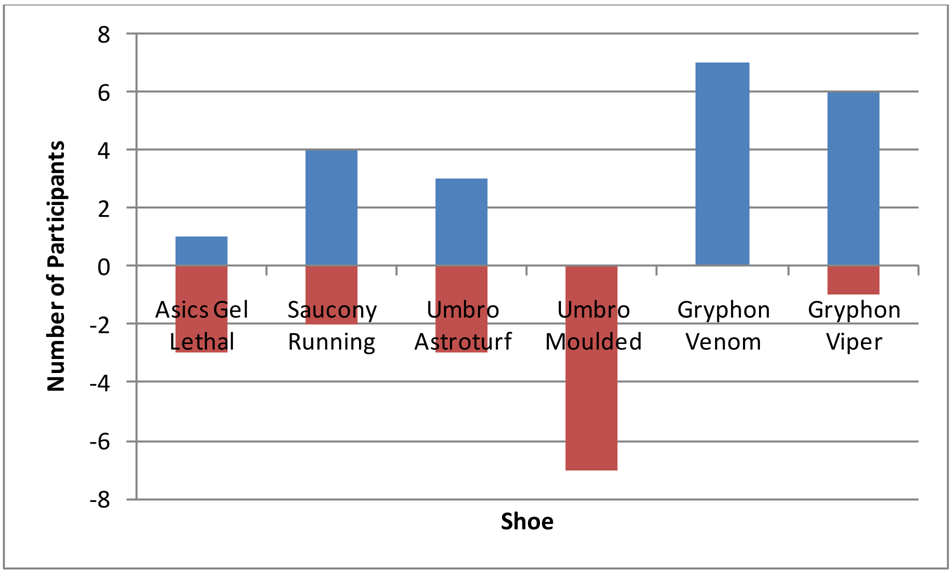
a) Time to 1st vertical GRF peak from foot down (TVFP1) while jogging



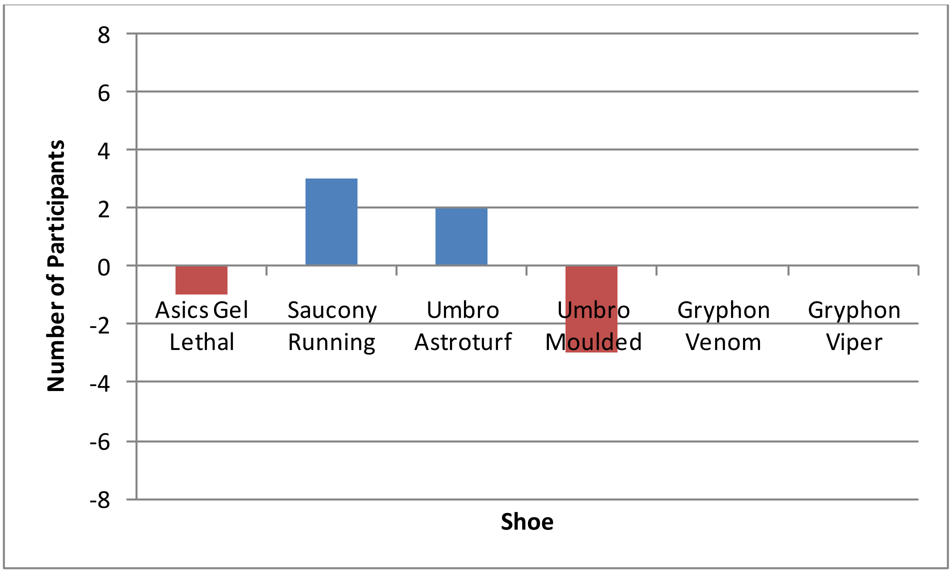
b) Time to 1st vertical GRF peak from foot down (TVFP1) while running

Figure 8.7 Total number of participants reporting significant differences (P<0.05) in the Vertical GRF 1st Peak, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower Force, Red (-ve) = Higher Force[7]

[7] For each participant investigated, each footwear design may have reported a significantly shorter time to 1st vertical GRF peak compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



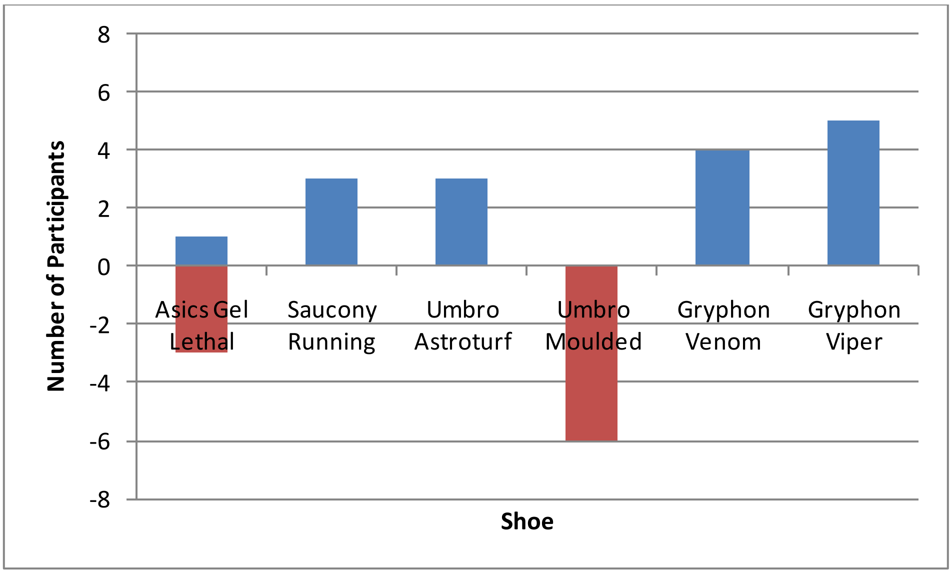
a) Time to 2nd vertical GRF peak from foot down (TVFP2) while jogging



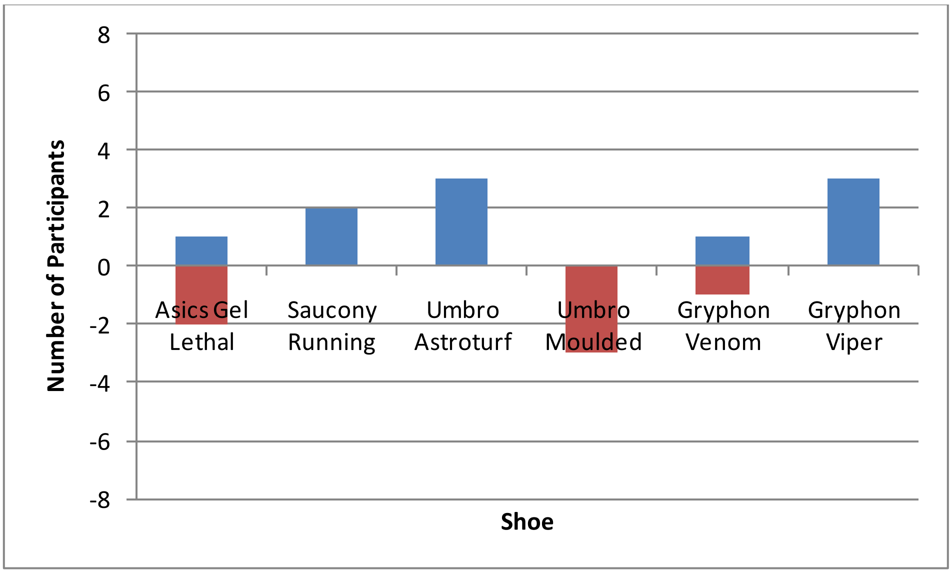
b) Time to 2nd vertical GRF peak from foot down (TVFP2) while running

Figure 8.8 Total number of participants reporting significant differences (P<0.05) in the time to vertical GRF 2nd Peak from foot down, when comparing the effects of each footwear to all other pairs. Blue (+ve) = longer period of time, Red (-ve) = Shorter period of time[8]

[8] For each participant investigated, each footwear design may have reported a significantly shorter time to 2nd vertical GRF peak compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



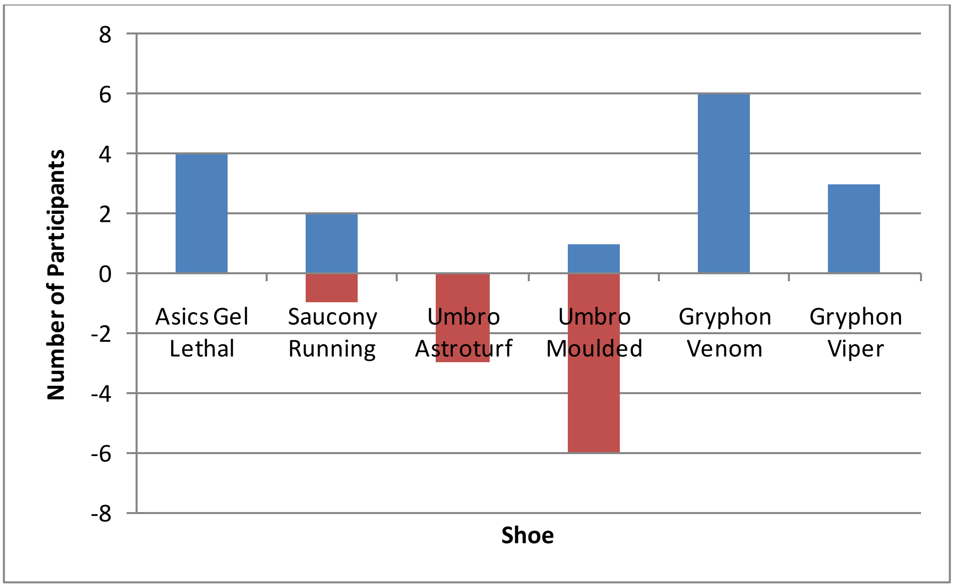
a) Time to peak vertical loading rate from foot down (TPVL) while jogging



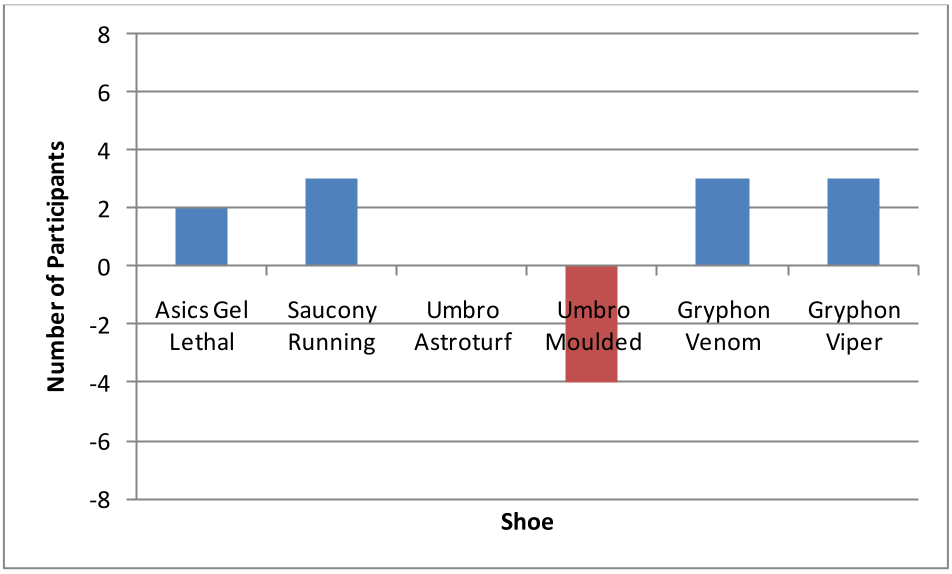
b) Time to peak vertical loading rate from foot down (TPVL) while running

Figure 8.9 Total number of participants reporting significant differences (P<0.05) in the time to peak vertical loading rate from foot down, when comparing the effects of each footwear to all other pairs. Blue (+ve) = longer period of time, Red (-ve) = Shorter period of time[9]

[9] For each participant investigated, each footwear design may have reported a significantly shorter time to 2nd vertical GRF peak compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



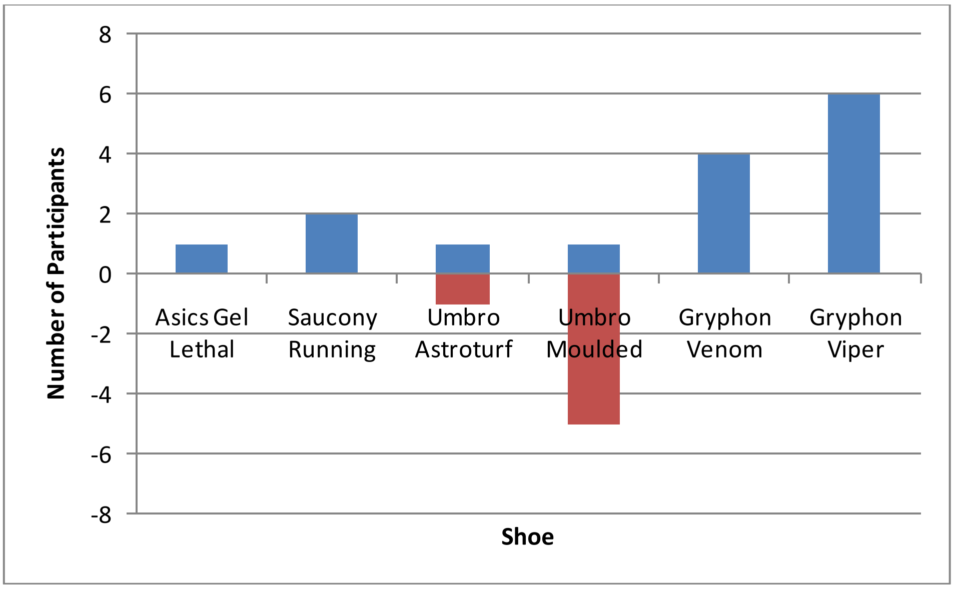
a) Average vertical loading rate to 1st vertical peak (AVLR) while jogging



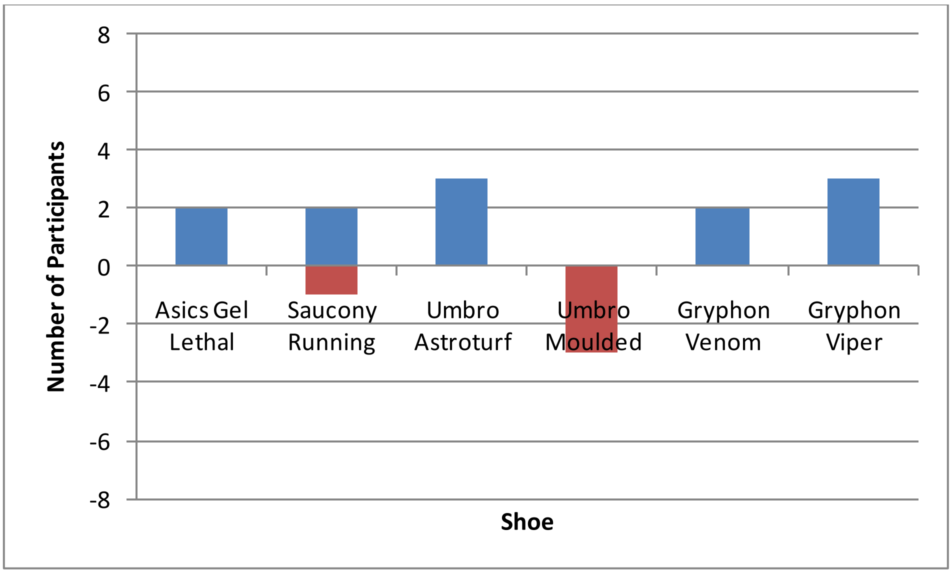
b) Average vertical loading rate to 1st vertical peak (AVLR) while running

Figure 8.10 Total number of participants reporting significant differences ($P<0.05$) in the average vertical loading rate to the 1st vertical force peak, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower loading rate, Red (-ve) = Higher loading rate[10]

[10] For each participant investigated, each footwear design may have reported a significantly higher AVLR compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



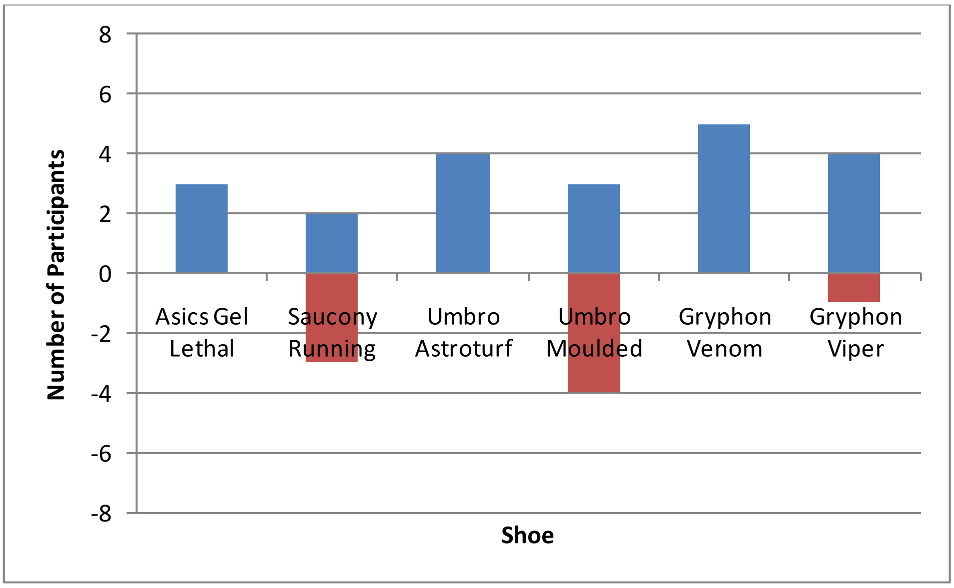
a) Average vertical loading rate from 50N to 50N Plus BW(AVL50NT50NBW) while jogging



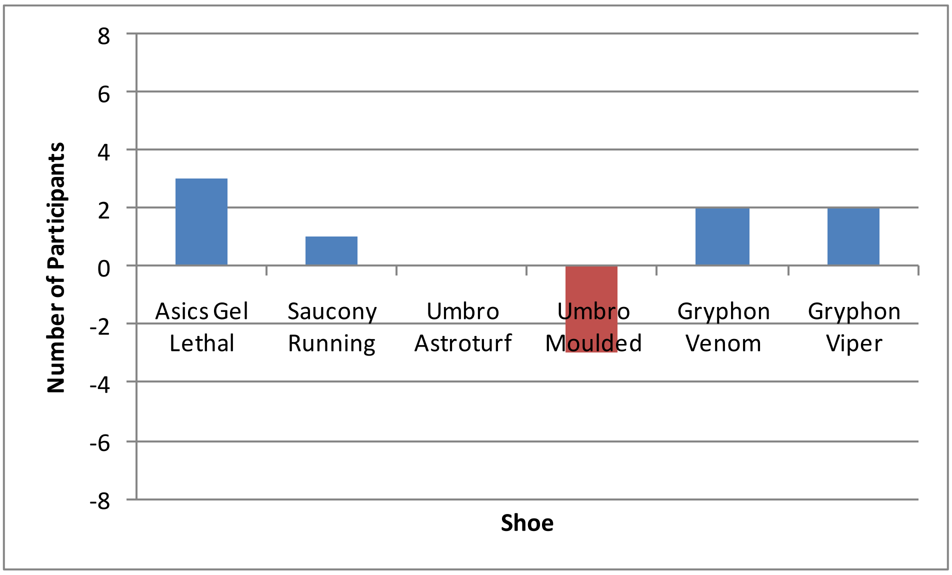
b) Average vertical loading rate from 50N to 50N plus BW (AVL50NT50NBW) while running

Figure 8.11 Total number of participants reporting significant differences (P<0.05) in the average vertical loading rate from foot 50N to 50N plus BW, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower loading rate, Red (-ve) = Higher loading rate.[11]

[11] For each participant investigated, each footwear design may have reported a significantly higher AVL50NT50NBW compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



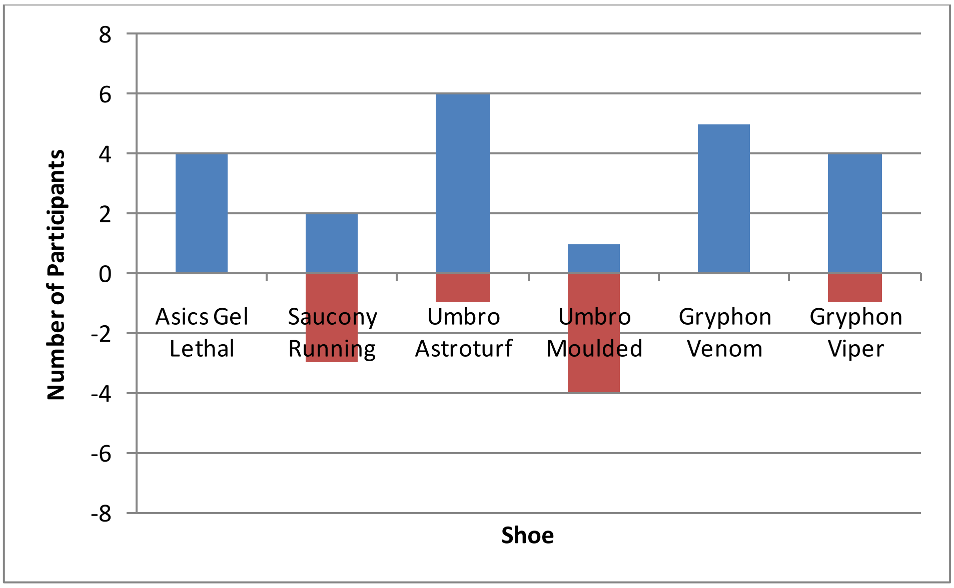
a) Average vertical loading rate from 20 to 90% of foot down to 1st vertical force peak (AVL20T90) while jogging



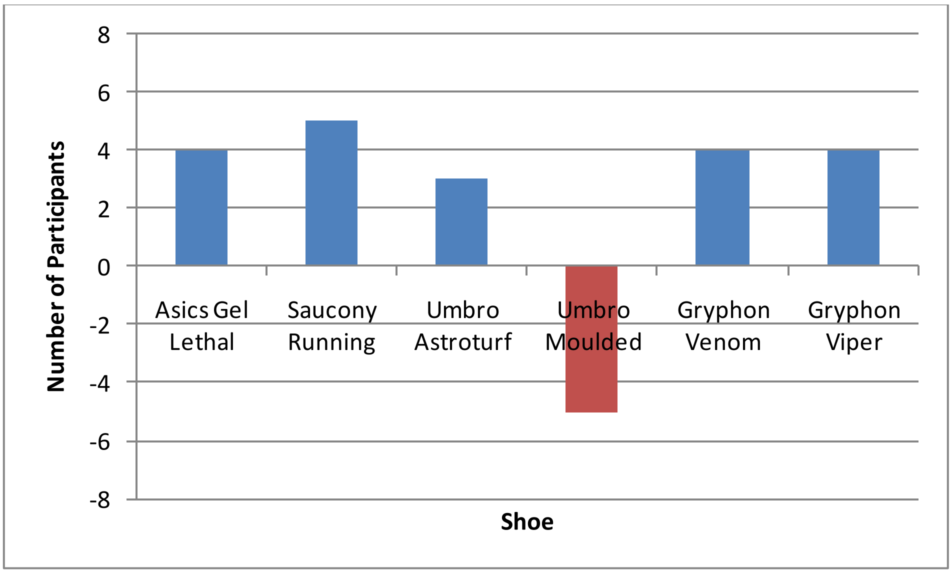
b) Average vertical loading rate from 20 to 90% of foot down to 1st vertical force peak (AVL20T90) while running

Figure 8.12 Total number of participants reporting significant differences (P<0.05) in the average vertical loading rate from 20 to 90% of the 1st vertical force peak, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower loading rate, Red (-ve) = Higher loading rate.[12]

[12] For each participant investigated, each footwear design may have reported a significantly higher AVL20T90 compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



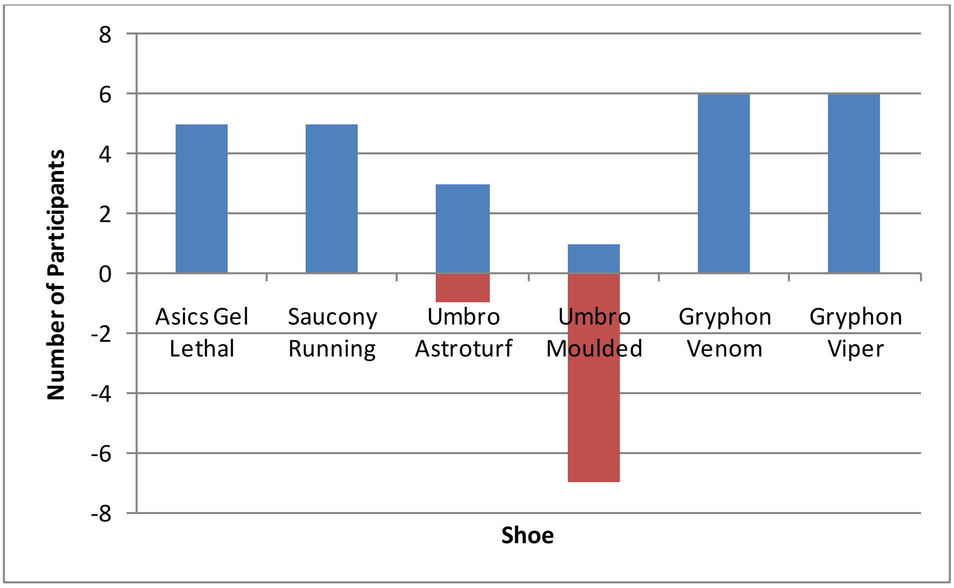
a) Average vertical loading rate from 20 to 80% of foot down to 1st vertical force peak (AVL20T80) while jogging.



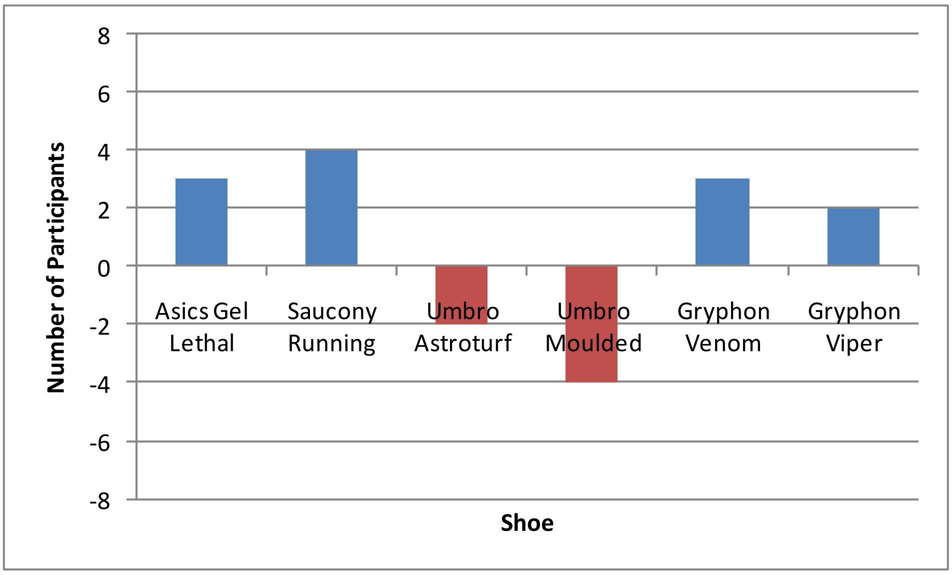
b) Average vertical loading rate from 20 to 80% of foot down to 1st vertical force peak (AVL20T80) while running.

Figure 8.13 Total number of participants reporting significant differences (P<0.05) in the average vertical loading rate from 20 to 80% of the 1st vertical force peak, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower loading rate, Red (-ve) = Higher loading rate.[13]

[13] For each participant investigated, each footwear design may have reported a significantly higher AVL20T80 compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)



a) Peak vertical loading rate (instantaneous) (PVLR) while jogging



b) Peak vertical loading rate (Instantaneous) (PVLR) while running

Figure 8.14 Total number of participants reporting significant differences (P<0.05) in the peak instantaneous loading rate, when comparing the effects of each footwear to all other pairs. Blue (+ve) = Lower loading rate, Red (-ve) = Higher loading rate[14]

[14] For each participant investigated, each footwear design may have reported a significantly higher PVLR compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 8 participant there can be a maximum of 8 positive and 8 negative outcomes (n=16)

This study has identified that different methods of measuring the GRF loading rate at impact affect the results of research. Significant differences were reported using PVLR and AVL50NT50NBW whereas no significant differences were reported for the AVLR, AVL20T90 and AVL20T80 variables. A reason for these differences may be due to the larger spread of the mean data identified in Figure 8.5 and 8.6. This may be due to having to identify the first GRF impact peak, which introduces human error. Furthermore, as the impact acceleration peak occurs prior to the GRF impact peak (Table 5.1), the value and timing of the impact peak should not be used in future when using GRF characteristics to identify changes in impact shock.

Table 8.2 Number of participants reporting significantly (P<0.05) Larger (L) and Smaller (S) GRF characteristics. Comparing (a) Asics Gel Lethal, (b) Saucony Running, (c) Umbro Astroturf, (d) Umbro Moulded, (e) Gryphon Venom and (d) Gryphon Viper shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

a) **Asics Gel Lethal**

		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper	
GRF Variable in Asics Gel Lethal shoes below compared to the shoes above:		S	L	S	L	S	L	S	L	S	L
TVFP1	Jogging on ISSS	1	0	0	0	0	1	2	0	1	0
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
TPVL	Jogging on ISSS	0	0	0	0	0	0	1	0	4	0
	Running on ISSS	0	0	2	0	0	0	0	0	3	0
VFP1	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
AVLR	Jogging on ISSS	0	0	0	0	5	0	0	0	0	0
	Running on ISSS	0	0	0	0	2	0	0	0	0	0
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	2	0	0	0	0	0
	Running on ISSS	0	0	0	0	2	0	0	0	0	0
AVL20T90	Jogging on ISSS	1	0	0	0	3	0	0	0	1	0
	Running on ISSS	0	0	0	0	3	0	0	0	0	0
AVL20T80	Jogging on ISSS	1	0	0	0	4	0	0	0	0	0
	Running on ISSS	0	0	0	0	3	0	0	0	0	0
PVLR	Jogging on ISSS	0	0	0	0	5	0	0	0	0	0
	Running on ISSS	0	0	4	0	3	0	0	0	0	0

b) **Saucony Running**

GRF Variable in Saucony Running shoes below compared to the shoes above:		Asics Gel Lethal S	Asics Gel Lethal L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L	Gryphon Viper S	Gryphon Viper L
TVFP1	Jogging on ISSS	0	1	0	1	0	4	2	0	1	0
	Running on ISSS	0	0	0	1	0	3	0	0	0	0
TPVL	Jogging on ISSS	0	0	0	0	0	2	0	0	0	0
	Running on ISSS	0	0	0	0	0	2	0	0	0	0
VFP1	Jogging on ISSS	0	0	0	1	0	2	0	0	0	1
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
AVLR	Jogging on ISSS	0	0	0	0	3	0	0	0	0	0
	Running on ISSS	0	0	0	0	3	0	0	0	0	0
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	3	0	0	0	0	0
	Running on ISSS	0	0	0	0	2	0	0	0	0	0
AVL20T90	Jogging on ISSS	0	1	0	2	4	0	0	1	0	0
	Running on ISSS	0	0	0	0	1	0	0	0	0	0
AVL20T80	Jogging on ISSS	0	1	0	2	3	0	0	2	0	0
	Running on ISSS	0	0	0	0	5	0	0	0	0	0
PVLR	Jogging on ISSS	0	0	1	0	5	0	0	0	0	0
	Running on ISSS	0	0	2	0	4	0	0	0	0	0

c) **Umbro Astroturf**

GRF Variable in Umbro Astroturf shoes below compared to the shoes above:		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L	Gryphon Viper S	Gryphon Viper L
TVFP1	Jogging on ISSS	0	0	1	0	0	3	2	0	1	0
	Running on ISSS	0	0	1	0	2	0	0	0	0	0
TPVL	Jogging on ISSS	0	0	0	0	0	3	0	0	0	0
	Running on ISSS	0	2	0	0	0	3	0	0	0	0
VFP1	Jogging on ISSS	0	0	1	0	0	1	0	0	0	0
	Running on ISSS	0	0	0	0	0	2	0	1	0	0
AVLR	Jogging on ISSS	0	0	0	0	0	1	0	1	0	1
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	2	1	0	1	0	1
	Running on ISSS	0	0	1	0	2	0	0	0	0	0
AVL20T90	Jogging on ISSS	0	0	2	0	0	0	0	0	0	1
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
AVL20T80	Jogging on ISSS	0	0	2	0	3	1	0	0	0	1
	Running on ISSS	0	0	0	0	3	0	0	0	0	0
PVLR	Jogging on ISSS	0	0	0	1	2	1	0	0	0	0
	Running on ISSS	0	0	0	2	0	0	0	1	0	0

d) **Umbro Moulded**

		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Gryphon Venom		Gryphon Viper	
GRF Variable in Umbro Moulded shoes below compared to the shoes above:		S	L	S	L	S	L	S	L	S	L
TVFP1	Jogging on ISSS	1	0	4	0	2	0	6	0	6	0
	Running on ISSS	0	0	3	0	2	0	1	0	2	0
TPVL	Jogging on ISSS	0	0	2	0	3	0	3	0	5	0
	Running on ISSS	1	0	2	0	3	0	1	0	3	0
VFP1	Jogging on ISSS	0	0	2	0	1	0	0	0	0	0
	Running on ISSS	0	0	0	0	2	0	0	0	0	0
AVLR	Jogging on ISSS	0	5	0	3	1	0	0	6	0	4
	Running on ISSS	0	2	0	3	0	0	0	3	0	3
AVL50NT50NBW	Jogging on ISSS	0	2	0	3	0	2	0	4	0	5
	Running on ISSS	0	2	0	2	0	2	0	2	0	2
AVL20T90	Jogging on ISSS	0	3	2	2	0	0	0	4	0	5
	Running on ISSS	0	2	0	0	0	0	0	2	0	2
AVL20T80	Jogging on ISSS	0	4	0	3	1	3	0	4	0	5
	Running on ISSS	0	4	0	5	0	3	0	4	0	4
PVLR	Jogging on ISSS	0	0	0	1	1	2	0	0	0	0
	Running on ISSS	0	3	0	4	0	0	0	3	0	2

e) **Gryphon Venom**

		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Viper	
GRF Variable in Gryphon Venom shoes below compared to the shoes above:		S	L	S	L	S	L	S	L	S	L
TVFP1	Jogging on ISSS	0	2	0	2	0	2	0	6	0	1
	Running on ISSS	0	0	0	0	0	0	0	1	0	0
TPVL	Jogging on ISSS	0	1	0	0	0	0	0	3	0	0
	Running on ISSS	0	0	0	0	0	0	0	1	1	0
VFP1	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0
	Running on ISSS	0	0	0	0	1	0	0	0	0	0
AVLR	Jogging on ISSS	0	0	1	0	6	0	0	0	0	0
	Running on ISSS	0	0	0	0	0	0	3	0	0	0
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	1	0	4	0	0	0
	Running on ISSS	0	0	0	0	0	0	2	0	0	0
AVL20T90	Jogging on ISSS	0	0	2	0	0	0	4	0	1	0
	Running on ISSS	0	0	0	0	0	0	2	0	0	0
AVL20T80	Jogging on ISSS	0	0	2	0	0	0	4	0	1	0
	Running on ISSS	0	0	0	0	0	0	4	0	0	0
PVLR	Jogging on ISSS	0	0	0	0	0	0	6	0	0	0
	Running on ISSS	0	0	0	0	0	0	3	0	0	0

f) **Gryphon Viper**

GRF Variable in Gryphon Viper shoes below compared to the shoes above:		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L
TVFP1	Jogging on ISSS	0	1	0	1	0	1	0	6	1	0
	Running on ISSS	0	0	0	0	0	0	0	2	0	0
TPVL	Jogging on ISSS	0	4	0	0	0	0	0	5	0	0
	Running on ISSS	0	3	0	0	0	0	0	3	0	1
VFP1	Jogging on ISSS	0	0	1	0	0	0	0	0	0	0
	Running on ISSS	0	0	0	0	0	0	0	0	0	0
AVLR	Jogging on ISSS	0	0	0	0	1	0	4	0	0	0
	Running on ISSS	0	0	0	0	0	0	3	0	0	0
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	0	0	5	0	0	0
	Running on ISSS	0	0	0	0	0	0	3	0	0	0
AVL20T90	Jogging on ISSS	0	1	0	0	1	0	5	0	0	1
	Running on ISSS	0	0	0	0	0	0	2	0	0	0
AVL20T80	Jogging on ISSS	0	0	0	0	1	0	5	0	1	0
	Running on ISSS	0	0	0	0	0	0	4	0	0	0
PVLR	Jogging on ISSS	0	0	0	0	0	0	7	0	0	0
	Running on ISSS	0	0	0	0	0	0	2	0	0	0

8.4 Conclusion

The results of this study conclude that identification of differences between conditions may best be identified using loading rate characteristics that do not require the identification of an impact peak, which is sometimes not present in data. Variables such as the PVLR or AVL50NT50NBW should therefore be used in future. Furthermore, timing variables such as TPVLR can also provide data that does not require identification of a peak and have been shown to be a useful variable within this study.

Considering these types of variables, the choice of footwear across a population of field hockey participants can influence the occurrence of injury. Evidence from this study suggests that by selecting the Saucony Running, Gryphon Venom, Gryphon Viper and Umbro Astroturf shoes instead of the Umbro Moulded shoes, detrimental loading linked to overuse injuries will be reduced across a population.

This study has also produced tables that can help assist in participants making informed choices on footwear (Table 8.2). By considering how many of the participants found favourable or unfavourable significant differences in GRF characteristics, an informed choice can be made as which footwear is likely to be the most effective in overuse injury prevention for an individual. However these tables are only a guide from the results of the eight participants' experiences in this study, which have demonstrated that shoes affect individuals differently. Therefore the research suggests that an individual should be assessed using the methods in this study to identify the correct footwear choice.

Chapter 9

Effects of Footwear on Tibial Accelerations

9 Effects of Footwear on Tibial Accelerations

9.1 Introduction

During human locomotion the body is exposed to GRFs causing musculoskeletal transient shocks that have been linked to various injuries (Snel *et al.*, 1985;Zhang *et al.*, 2008;Milner *et al.*, 2006;Verbitsky *et al.*, 1998). The relationship between GRFs and Impact shock has been measured in this thesis (Chapter 5). The rates of the loading have been identified as being correlated to peak impact shocks through the tibia.

The magnitude of the GRFs and impact shocks have also been reported as being affected by the velocity of the locomotion (Perry and Lafortune, 1995;Weyand *et al.*, 2000), movement strategies (Oakley and Pratt, 1988;Laughton *et al.*, 2003;Lieberman *et al.*, 2010), the surface (Riley *et al.*, 2007;Riley *et al.*, 2008;Hardin *et al.*, 2004;Dixon *et al.*, 2000;Stussi *et al.*, 1997) and the footwear worn (McNair and Marshall, 1994;Clarke *et al.*, 1983a;Aguinaldo and Mahar, 2003). The velocity and intensity of locomotion during training and matches can be controlled by an individual player. However if a player is attempting to perform to the best of their ability, restricting their movement to reduce detrimental GRFs is not desirable or realistic in practice. Participants also have limited control over the surface on which they play although during warm ups and training there may be an option to change surfaces. During warm up periods before matches due to pitch availabilities and costs, teams will often warm up off the pitches on harder concrete surfaces. This is often done in the same footwear as which they are about to play the match. Various footwear designs are used by participants at club through to elite level. Participants choose footwear for various reasons such as comfort, grip characteristics, protection from side impacts to the uppers of the feet, weight, and the way the footwear looks. Footwear is one of the factors affecting impact shock attenuation that an individual player has an individual choice over.

A new running shoe (Healus® Running Shoe) that is designed with the heel section removed is being developed (Figure 9.1). This shoe is currently not available on the market however access to a proto type of this shoe was given for this research by the manufacturer. This shoe may be an effective new design in training footwear that could be used by field hockey to reduce the occurrence of overuse injuries. The shoe is introduced at this point in this thesis, as it was only available from the manufacturer at the time of collecting the data for this final study.

Figure 9.1 Healus® Running Shoes

The Healus® Running shoe is designed to encourage a more forefoot landing movement in athletes. Previous research has identified forefoot landing compared to rearfoot landing strategies as reducing impacts shock measured with accelerometers attached to the human body and loading rates linked to impact shock (Lieberman *et al.*, 2010;Oakley and Pratt, 1988;Arendse *et al.*, 2004). However in shod conditions without sufficient training, an increase in impact accelerations has been reported (Laughton *et al.*, 2003). This new shoe is designed to alter the movement strategy without the athlete having to consciously change how they run, as in the previously mentioned research. Therefore a reduction in the magnitude of impact accelerations may be achievable without extensive previous training. This would allow athletes to use such a shoe during training to reduce the risk of overuse injury. However the shoes may expose athletes to a higher risk of other injuries due to a possible reduction in stability with the presence of a relatively thick midsole. However previous research in shoes with increased thick mid-sole cushioning systems did not identify any increases in ankle sprains during basketball participation (Curtis *et al.*, 2008). This may be an area for future research for this particular design of footwear.

9.1.1 Measuring Tibial shock

Previous research earlier in this thesis (Chapter 5) investigated the links between accelerometers mounted to the skin at the posterior medial aspect of the shank and various GRF characteristics. Moderate links were reported between various methods of analysing loading rates and the peak accelerations recorded. However other research has reported much higher correlations between loading rates and tibial impact shock recorded from bone mounted and skin mounted accelerometers (Hennig and Lafortune, 1991;Laughton *et al.*, 2003;Hennig *et al.*, 1993). Tibial shock can be measured more accurately by attaching accelerometers invasively via a pin, directly to the bone (Lafortune et al., 1995a). This is the most direct approach to measuring tibial acceleration however it is not always practical due to the invasive nature of this methodology. Mounting accelerometers over the skin at bony points of the body provides a more practical alternative. However, the acceleration signal from skin mounted accelerometers has been reported as on average to be twice the magnitude of bone mounted systems (Lafortune *et al.*, 1995a). Using a Butterworth low pass filter the unwanted high frequency signal components due to the skin interaction can be reduced (Shorten and Winslow, 1992). Measuring tibial accelerations directly by attaching an accelerometer to skin covering the medial posterior aspect of the tibia has since become an accepted methodology for measuring impact shock (Coventry *et al.*, 2006;Flynn *et al.*, 2004;Pohl *et al.*, 2008;Laughton *et al.*, 2003;Milner *et al.*, 2006).

9.2 Methodology

Nine field hockey participants, all adult males (Age 21 $\pm$ 1.69, Height 175.75 $\pm$ 6.56 and Mass 78.13 $\pm$ 12.11) volunteered to take part in this study. All participants were injury free at the time of data collection and completed an informed consent form.

The same accelerometer attachment methodology was used in this study as was used in the investigation in chapter 5. A tri-axial accelerometer (Biometrics ACL300) was mounted to a lightweight carbon-fibre plate via a securely glued lightweight bolt and thread attachment. The total weight of the accelerometer and mounting system was 13g. The carbon-fibre plate was securely attached to the shank via surgical adhesive tape. By using skin stretching techniques the plate was attached tightly so the accelerometer was positioned on the distal anterio-medial aspect of the tibia and 8cm above the medial-malleolus. The accelerometer was orientated to measure the acceleration in the direction along the longitudinal axis of the tibia (Figure 9.2). By positioning the accelerometer near the malleolus, the radius of the motion of the sensor about the ankle joint was minimised thus reducing the effects of the centripetal accelerations resulting from the angular motion highlighted in previous research (Laughton *et al.*, 2003). The accelerometer system used had an inbuilt low frequency filter that excluded the proportion of the signal due to gravity. The voltage sensitivity of the accelerometer (maximum 1000g measurement) signal was set to 100mV/g, allowing adequate sensitivity with a measurement range of ±100g and the sampling frequency was set to 1000Hz. The accelerometer analogue signal was recorded by a Biometrics DataLog system (Biometrics Ltd, Gwent, UK) securely fastened to the participant via a back pack. This allowed the participants to be free moving and did not require them to land their foot in any specific areas allowing for a more natural movement than in many studies.

Figure 9.2 Accelerometer attached to the antero-medial aspect of the shank

Two sets of timing gates were positioned 10m apart and on the edge of a synthetic sports surface at Preston Sports Arena so that there was a runway between the timing gates on the synthetic surface and the concrete surface at the side of the pitch. The concrete surface at the side of the pitch was typical of the sort of surface on which field hockey participants would warm up prior to matches as the pitch would often be occupied up until just before the start of the match. Participants were required to run in each of the footwear at 5m.s^{-1} (running) and 3.3m.s^{-1} (jogging), between the two

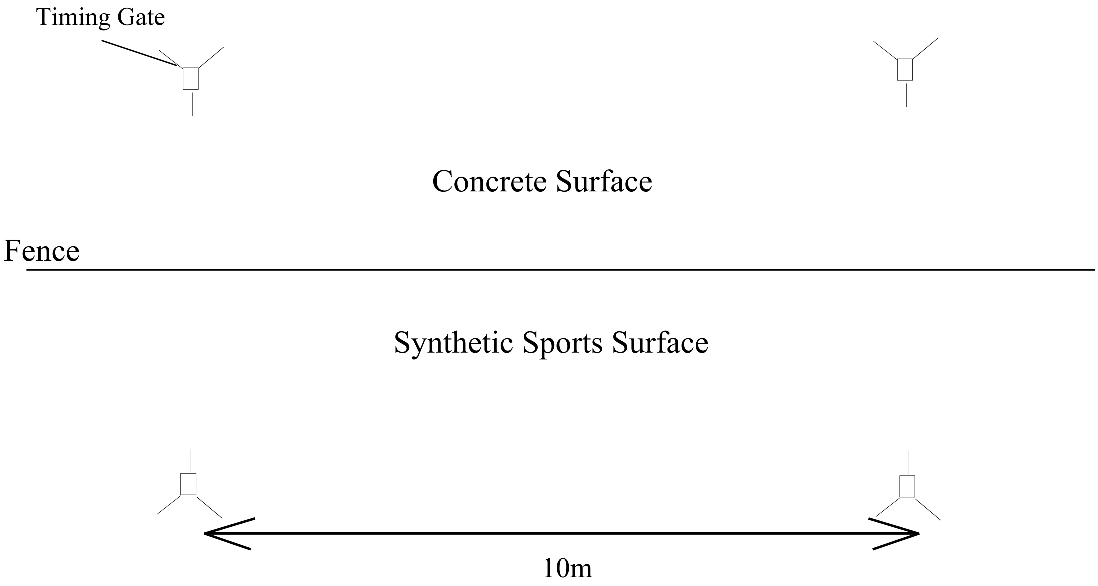


Figure 9.3 Setup of timing gates to cover locomotion on concrete and the synthetic sports surface

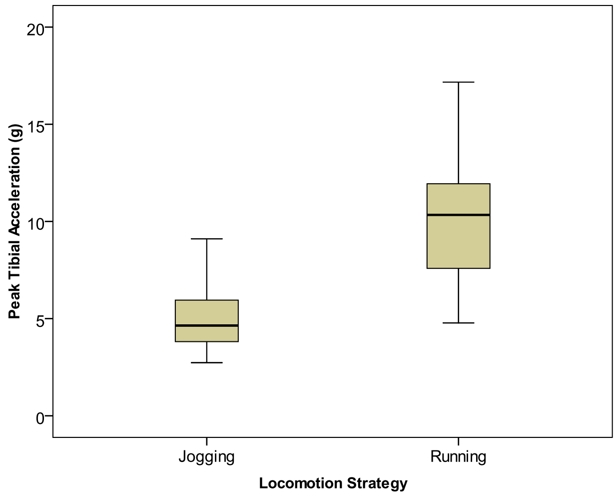
sets of gates twice on each surface. Data would be considered good if the participant ran within 5% of the desired running speeds. For a good trial 3 foot to ground impact accelerations were recorded by the shank mounted accelerometer. By pacing out from the participants' starting point which was approximately 10 metres to allow them sufficient room to accelerate, the number of steps was noted to allow identification of the three foot to ground impacts that were recorded in between the timing gates. By recording two sets of three impacts for each condition (footwear x surface x speed) six trials would be recorded for nine participants which fits in the scale identified by Bates and colleagues of trials to sample size for sufficient statistical power for participants using similar performance strategies (Bates *et al.*, 1992). The accelerometer signal was processed through a Butterworth low-pass filter set to 60Hz to exclude the component of the signal due to skin artefact and the resonance of the device in line with the findings from previous research (Shorten and Winslow, 1992).

9.3 Results and discussion

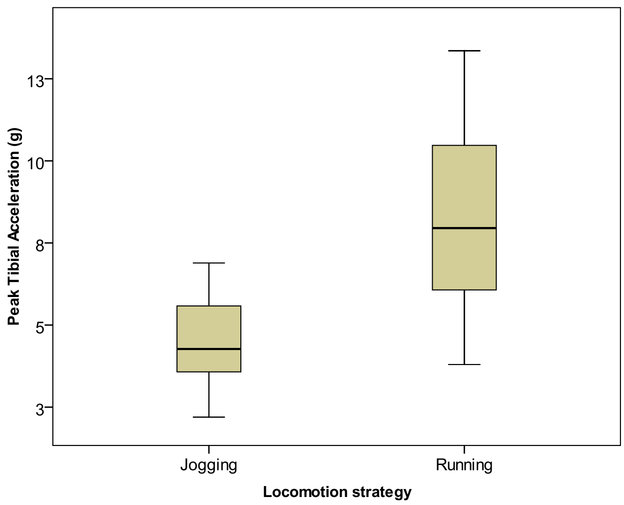
ANOVAs were performed to compare the means of all the participants' peak accelerations measured between the surfaces and the speed conditions. Post hoc analysis using a bonferonni test found significant differences ($P<0.05$) between running and jogging for the different surfaces, for data collected in all the different shoes (excluding the Umbro Moulded shoes) (Figure 9.4). Mean peak tibial axial accelerations of values of 9.8 and 8.3g were recorded for mean running values on concrete and the synthetic sports surface respectively, while smaller values of 5.1 and 4.8g were recorded during jogging on the same surfaces. Similar increases in tibial shock linked to overuse injury have been reported previously (Perry and Lafortune, 1995). It clearly demonstrates that higher velocity locomotion exposes the musculoskeletal system to larger impact transient shocks that would increase the risk of suffering a tibial stress fracture. Significant differences in the tibial axial peak were also reported when comparing the effects of surface during running (Figure 9.5a), however during jogging significant differences were not found (Figure 9.5b). These results show that such increases in velocity of locomotion (from 3.3 to $5 m.s^{-1}$) have a greater influence on the magnitude of impact shocks than surfaces typically used in field hockey. The results also show that jogging on a harder surface may not expose the participants to any significant increase in sustaining an overuse injury linked to higher levels of impact shock. However, during higher velocity activities this is not the case.

Mean and standard deviations of tibial accelerations for all footwear conditions are reported in Table 9.1. ANOVAs were performed to compare the means of all participants' peak accelerations at the tibia. Post hoc analysis using a bonferonni test found significant differences ($P<0.05$) between mean tibial acceleration values in the Saucony Running (8.0 ±2.8g), Umbro Soccer Astroturf (7.4 ±2.6g), and Gryphon Venom (7.9 ±3.0g), when compared to Umbro Soccer Moulded (13.1 ±4.4g) during running on synthetic surfaces. No other significant differences were found across the population of participants between the shoes for the 3 other surface and velocity conditions (running on synthetic surface, jogging on concrete, running on concrete).

From these results it would appear that footwear only has a measureable effect across the population of participants, during running at a higher speed ($5m.s^{-1}$). The significant differences recorded were all cases where the Umbro Moulded shoes exposed the athlete to higher tibial axial accelerations than the other shoe conditions. This occurred only in the running group on the synthetic sports surface This may have been replicated on the concrete but it was considered not safe for participants to run in Umbro Moulded shoes on a concrete surface. These findings are similar to the results from the previous chapter, where significantly different loading rates were reported between footwear conditions during running only. Furthermore, the same footwear (Umbro Moulded) as in the previous chapter was highlighted as exposing the population to a significant increase in detrimental kinetic factors.

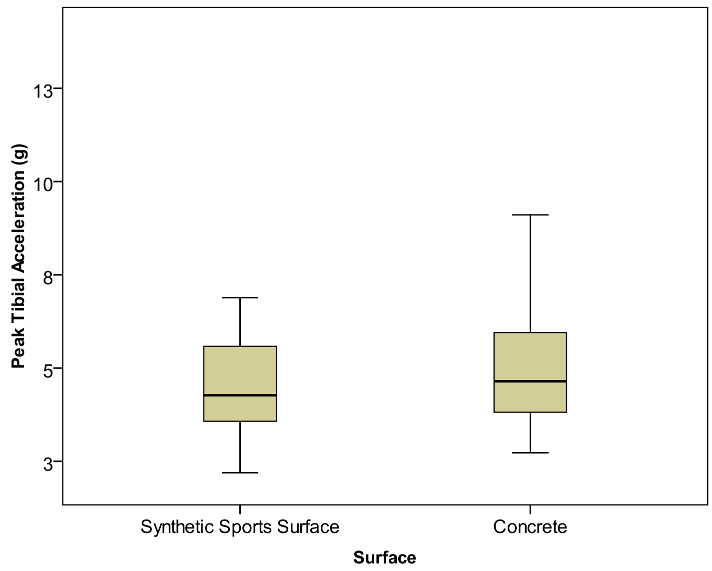


a) Concrete surface

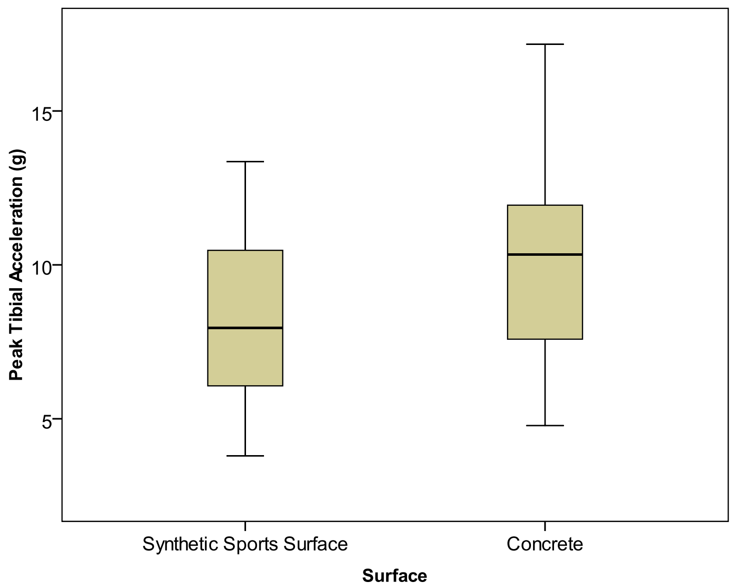


b) Synthetic sports surface

Figure 9.4 Peak mean tibial axial accelerations from all shoes and participants data comparing, a) Locomotion strategy on a concrete surface, b) Locomotion strategy on a synthetic sports surface



a) Jogging



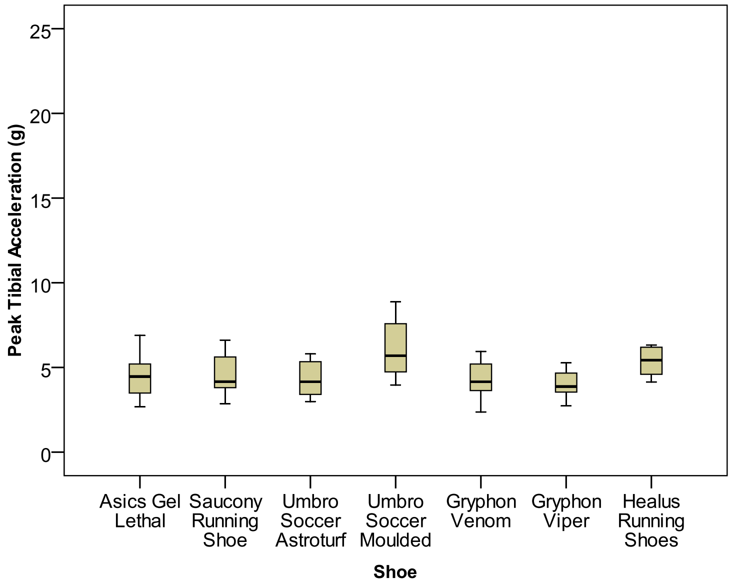
b) Running

Figure 9.5 Peak mean tibial axial accelerations from all shoes and participants data comparing, a) Surfaces whilst Jogging, b) Surfaces whilst running

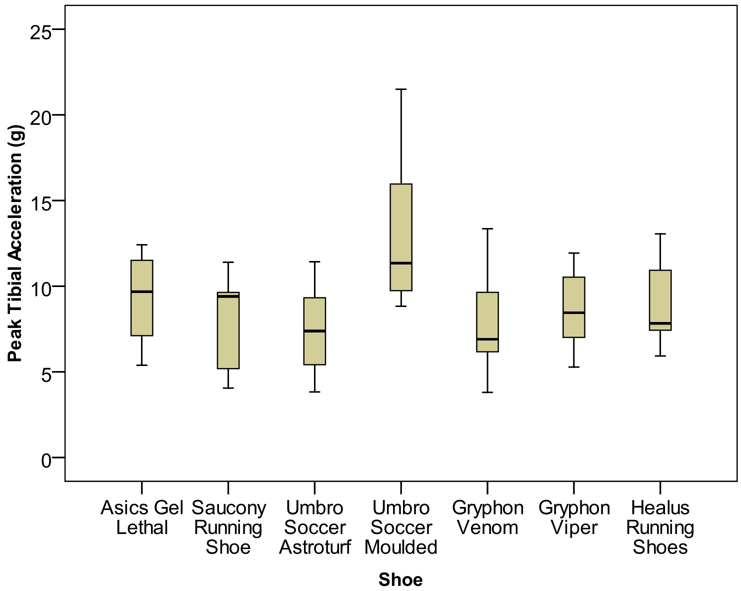
Table 9.1 Mean Peak Tibial Accelerations for all Participants

Footwear	Locomotion Condition	Peak Tibial Acceleration (g)	
Asics Gel Lethal	Jogging on OSSS	4.5	±1.4
	Running on OSSS	9.1	±2.7
	Jogging on Concrete	5.5	±1.8
	Running on Concrete	10.3	±2.4
Saucony Running	Jogging on OSSS	4.9	±1.9
	Running on OSSS	8.0	±2.8
	Jogging on Concrete	4.8	±1.8
	Running on Concrete	8.4	±2.7
Umbro Soccer Astroturf	Jogging on OSSS	4.8	±2.1
	Running on OSSS	7.4	±2.6
	Jogging on Concrete	4.8	±2.1
	Running on Concrete	10.2	±3.7
Umbro Soccer Moulded	Jogging on OSSS	6.6	±2.9
	Running on OSSS	13.1	±4.4
Gryphon Venom	Jogging on OSSS	5.0	±3.2
	Running on OSSS	7.9	±3.0
	Jogging on Concrete	5.0	±3.0
	Running on Concrete	10.4	±3.6
Gryphon Viper	Jogging on OSSS	4.2	±1.2
	Running on OSSS	8.6	±2.3
	Jogging on Concrete	4.9	±1.7
	Running on Concrete	10.5	±2.4
Healus® Running	Jogging on OSSS	5.4	±1.8
	Running on OSSS	8.9	±2.6
	Jogging on Concrete	5.6	±1.5
	Running on Concrete	9.3	±2.4

Comparing the effects of the footwear on each participant individually, demonstrated that any footwear could be identified as having significantly higher or lower impact accelerations compared to any of the other footwear (Figures 9.8 and 9.9). The results give further evidence that the Umbro Moulded shoes offer the least protection against high magnitude impact shocks. With nearly all participants experiencing significant increase in tibial accelerations compared to at least one of the other footwear. In general the other shoes performed similarly in the various conditions.

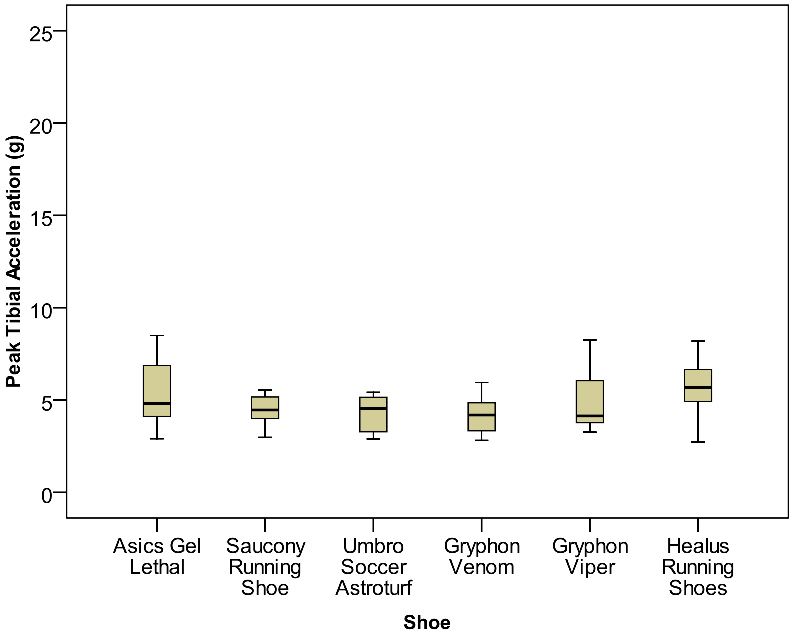


a) Jogging on Synthetic Surface

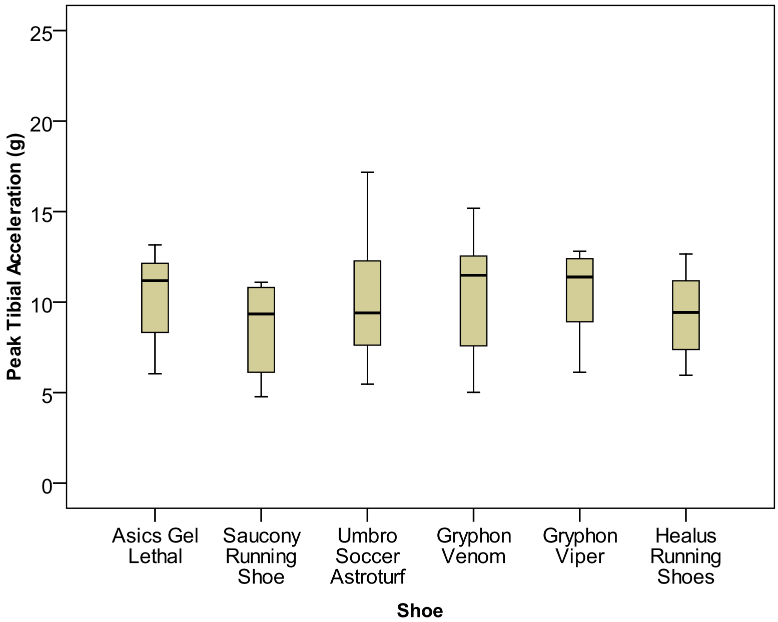


b) Running on Synthetic Surface

Figure 9.6 Peak mean tibial axial accelerations and high to low peak ranges for different shoes worn during a) Jogging on Synthetic Surface and b) Running on Synthetic Surface

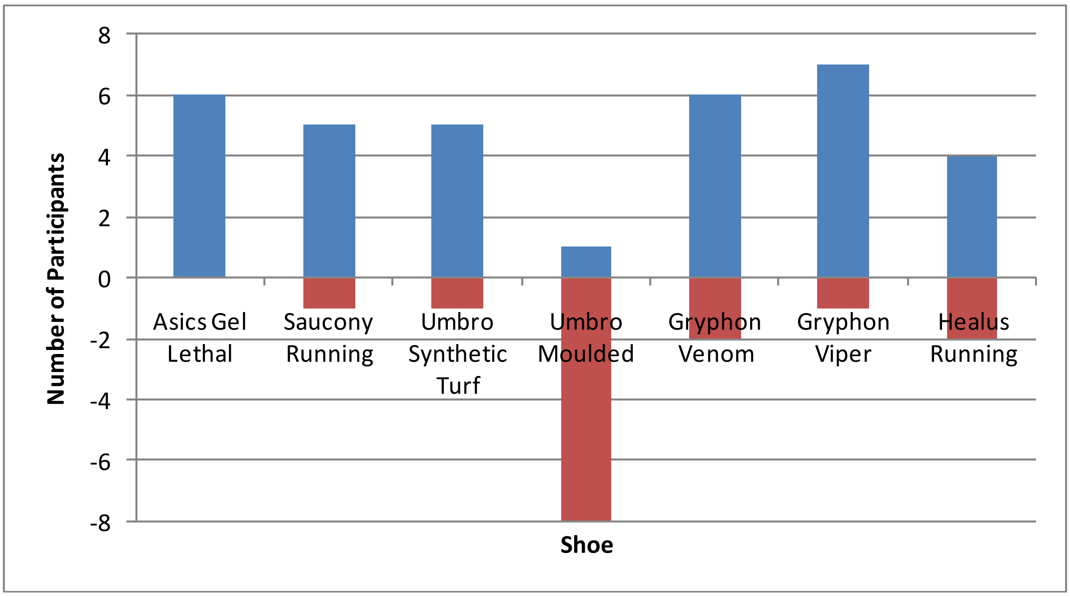


a) Jogging on Concrete

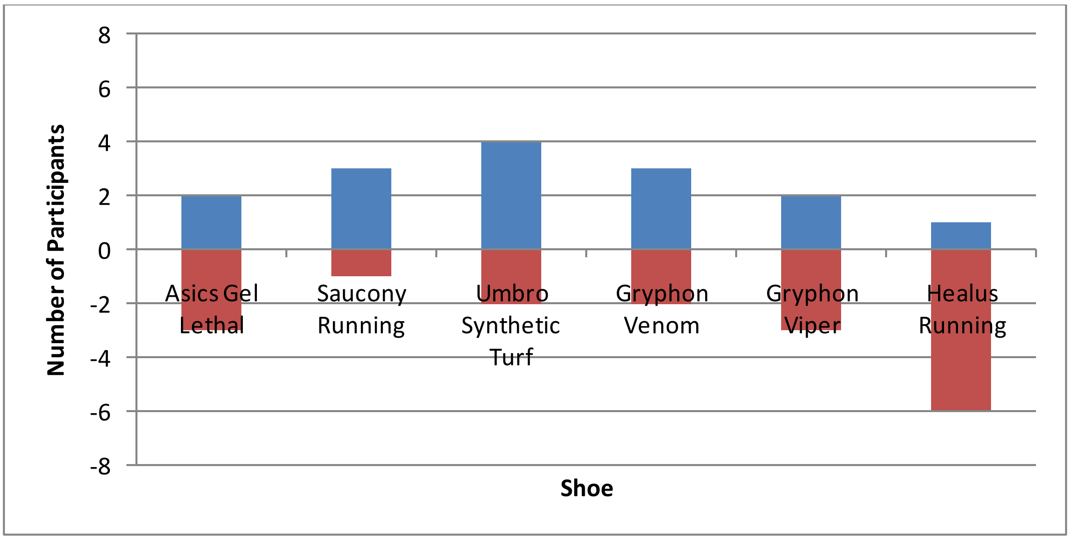


b) Running on Concrete

Figure 9.7 Peak mean tibial axial accelerations and high to low peak ranges for different shoes worn during a) Jogging on Concrete and b) Running on Concrete



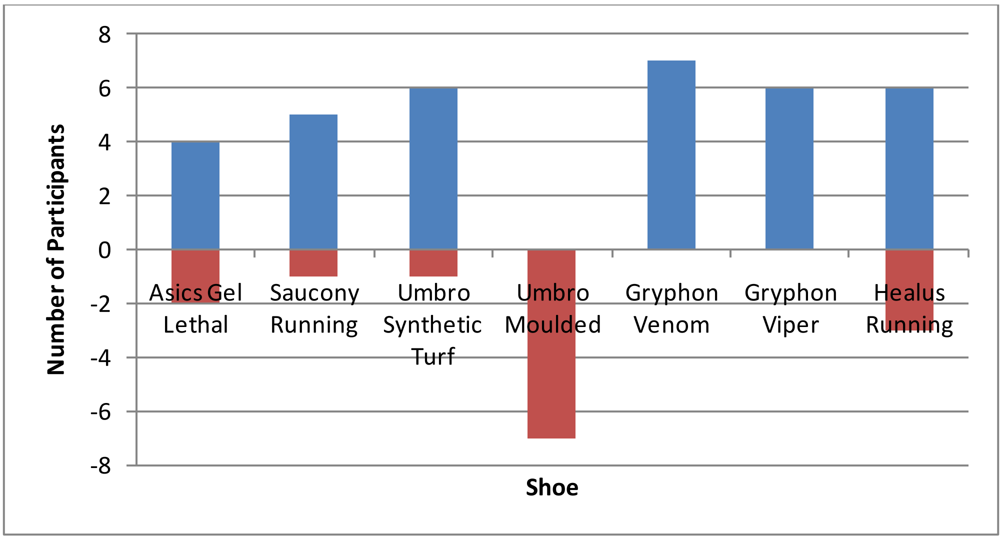
a) Jogging on a synthetic sports surface



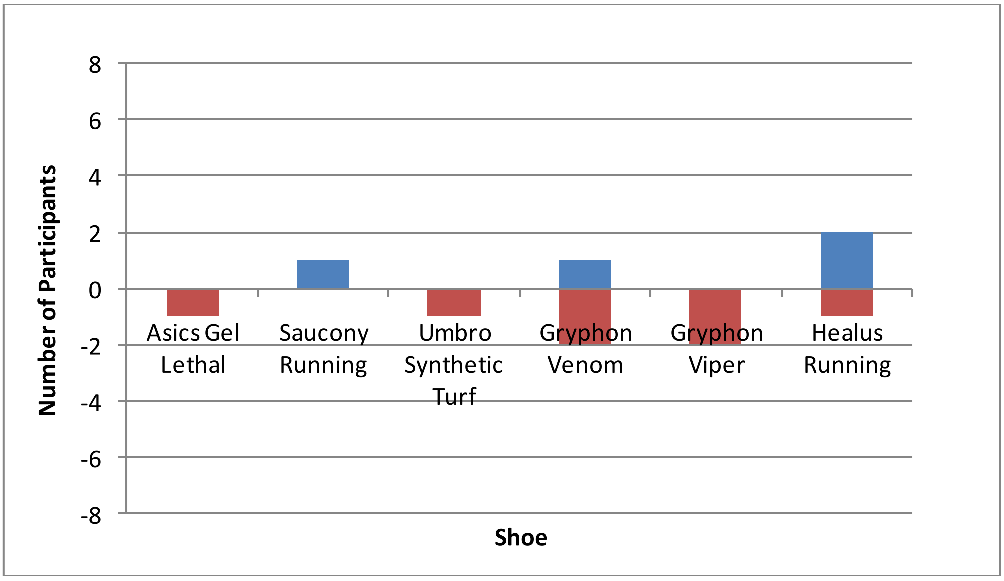
b) Jogging on concrete

Figure 9.8 Number of participants reporting significant differences in the magnitude of the tibial acceleration when comparing the effects of each footwear condition to all other pairs. Blue = lower tibial acceleration, Red = higher tibial acceleration.[15]

[15] For each participant investigated, each footwear design may have reported a significantly higher peak tibial acceleration compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 9 participants there can be a maximum of 9 higher and 9 lower tibial acceleration outcomes (n=18)



a) Running on a synthetic sports surface



b) Running on concrete

Figure 9.9 Number of participants reporting significant differences in the magnitude of the tibial acceleration when comparing the effects of each footwear condition to all other pairs. Blue = lower tibial acceleration, Red = higher tibial acceleration.[16]

[16] For each participant investigated, each footwear design may have reported a significantly higher peak tibial acceleration compared to another design of footwear, yet reported a longer value when compared to a further design of footwear. Therefore as there were 9 participants there can be a maximum of 9 higher and 9 lower tibial acceleration outcomes (n=18)

The Healus® Running shoes exposed 6 out of the 8 participants to significant (P>0.05) increases in tibial shock during jogging on concrete. This was reflected in the mean values across the population, with the Healus® Running shoe producing the highest peak acceleration. However, as explained earlier, this value was not significantly different to any of the other footwear tested.

When comparing each type of footwear to each of the other types individually, significant differences (P<0.05) were found. Table 9.2 uses the analysis methodology implemented in the last two chapters (Table 7.2 and 8.2). The effect of each shoe on the peak accelerations measured at the tibia is directly compared to each of the other shoes. If a significant difference within a participant's data is present, it is reported as either a larger or smaller peak acceleration. The data with smaller values is coloured red and data with larger values is coloured yellow. Therefore in general, for the shoe being investigated, the more red for a condition the more effective at reducing the impact shock the shoe was for more individuals in the group of participants. The more yellow and the shoe will have reported increases in the impact shock measured in more participants than those experiencing reductions. This method of presenting the results allows direct comparison between shoes. An important initial observation is that in some footwear comparisons, each of the footwear conditions are more effective and less effective than the other in reducing the impact accelerations for different participants. This shows that footwear which exposes a participant to lower impact accelerations can expose another participant to higher impact accelerations. Similar results to this were reported for the effects of surfaces on loading rates (highly correlated to impact shock) in individuals (Dixon *et al.*, 2000). The results of this study show that from the shoes investigated identifying a single shoe that would be best for any individual is not possible.

Table 9.2 Number of participants reporting significantly (P<0.05) Larger (L) and Smaller (S) peak tibial acceleration. Comparing (a) Asics Gel Lethal, (b) Saucony Running, (c) Umbro Astroturf, (d) Umbro Moulded, (e) Gryphon Venom, (f) Gryphon Viper and (g) Healus® Running shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

Asics Gel Lethal

		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus® Running	
Value of the Asics Gel Lethal Variable Below, Compared to the Condition Above		S	L	S	L	S	L	S	L	S	L	S	L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	0	6	0	1	0	0	0	2	0
	Jogging on Concrete	0	0	0	1			0	1	0	2	0	1
	Running on OSSS	0	2	1	2	4	0	1	1	0	0	1	0
	Running on Concrete	0	0	0	0			0	0	0	0	0	1

a) Saucony Running

		Asics Gel Lethal		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus ® Running	
Value of the Saucony Running Variable Below Compared to the Condition Above:		S	L	S	L	S	L	S	L	S	L	S	L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	1	3	0	1	1	0	1	2	0
	Jogging on Concrete	0	0	0	0			0	1	0	1	2	1
	Running on OSSS	2	0	0	1	5	0	0	0	0	0	2	0
	Running on Concrete	0	0	0	0			1	0	1	0	1	0

b) Umbro Astroturf

		Asics Gel Lethal		Saucony Running		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus ® Running	
Value of the Umbro Astroturf Variable Below, Compared to the Condition above:		S	L	S	L	S	L	S	L	S	L	S	L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	1	1	1	5	0	0	1	0	1	1	0
	Jogging on Concrete	1	0	0	0			0	1	0	1	1	1
	Running on OSSS	2	1	1	0	5	0	1	1	2	0	3	0
	Running on Concrete	0	0	0	0			0	1	0	0	0	1

c) Umbro Moulded

Value of the Umbro Moulded Variable below, Compared to the Condition above:		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Gryphon Venom S	Gryphon Venom L	Gryphon Viper S	Gryphon Viper L	Healus ® Running S	Healus ® Running L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	**0**	**6**	0	3	**0**	**5**	**0**	**6**	**0**	**6**	1	4
	Jogging on Concrete												
	Running on OSSS	**0**	**4**	**0**	**5**	**0**	**5**	**0**	**7**	**0**	**5**	**0**	**6**
	Running on Concrete												

d) Gryphon Venom

Value of the Gryphon Venom Variable Below, Compared to the Condition above:		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Viper S	Gryphon Viper L	Healus ® Running S	Healus ® Running L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	1	1	1	1	0	**7**	**0**	0	2	2	1
	Jogging on Concrete	1	0	1	0	1	0			0	0	2	0
	Running on OSSS	1	1	0	0	1	1	**7**	**0**	1	0	3	1
	Running on Concrete	0	0	0	1	1	0			0	0	0	1

e) Gryphon Viper

Value of the Gryphon Viper Variable Below, Compared to the Condition above:		Asics Gel Lethal S	Asics Gel Lethal L	Saucony Running S	Saucony Running L	Umbro Astroturf S	Umbro Astroturf L	Umbro Moulded S	Umbro Moulded L	Gryphon Venom S	Gryphon Venom L	Healus® Running S	Healus® Running L
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	0	1	0	**6**	**0**	2	0	2	1
	Jogging on Concrete	2	0	1	0	0	0			0	0	1	0
	Running on OSSS	0	0	0	0	0	2	**5**	**0**	0	1	2	0
	Running on Concrete	0	0	0	1	0	0			0	0	0	1

f) Healus Running

Value of the Healus® Running Variable Below, Compared to the Condition above:		Asics Gel Lethal L	Asics Gel Lethal H	Saucony Running L	Saucony Running H	Umbro Astroturf L	Umbro Astroturf H	Umbro Moulded L	Umbro Moulded H	Gryphon Venom L	Gryphon Venom H	Gryphon Viper L	Gryphon Viper H
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	2	0	2	0	1	4	1	1	2	1	2
	Jogging on Concrete	1	0	1	2	1	0	**6**	**0**	0	2	0	1
	Running on OSSS	0	1	0	2	0	3	1	3	0	2	0	2
	Running on Concrete	1	0	0	1	1	0	1	0	1	0	1	0

Comparing the results of kinetic footwear testing in this manner (Table 9.2) clearly identifies the Umbro Moulded as the most unfavourable shoe with at least 3 of the participants reporting significant differences compared to each of the other shoes. However, it should be noted that for one participant the shoes were favourable in jogging compared to the Saucony Running shoes. This highlights how the influence footwear choice can have is very dependent of the individual. The results of the other shoes are rather inconclusive however they do provide evidence that could be used for footwear choice. An example of this is that it would appear that the Healus® footwear would most likely be an unfavourable choice for activities on synthetic sports surfaces. The inconclusive results across direct footwear comparisons show a need for individuals to be biomechanically assessed if an informed choice is truly to be made.

The results from this study show that the moulded shoe design would expose field hockey participants to an increase in tibial shock and thus an increase in the risk of sustaining an overuse injury. The Healus® footwear would also appear to pose a risk during jogging on concrete. This may be due to participants not used to running in the shoes which attempt to adjust the movement strategy of the participants to a forefoot strike. While jogging the athletes may have been less comfortable in the shoes. Previous research has found that adjustment of movement strategies after extensive training can reduce loading and impact shock in shod conditions (Arendse *et al.*, 2004). However without suitable training adjustment when changing from heel strike to forefoot landing styles of shod running an increase in the magnitude of impact shock has been identified (Laughton *et al.*, 2003). Previous extensive training in this new design of footwear may produce more favourable results in reducing impact shock.

The influence of speed is a factor identified in this study that should be considered in injury prevention. If an athlete wishes to reduce their exposure to potentially damaging impact transient shocks then by reducing their speed they may achieve this regardless of surface. This agrees with previous research that identified speed as increasing the magnitude of impact shock (Perry and Lafortune, 1995). This may have little relevance during participation in sports competition. However during training and in particular for athletes prone to injuries relating to higher levels of impact shock, reducing the velocity of their locomotion where possible may help reduce the onset of overuse injuries.

When at relatively low speeds ($3.3 m.s^{-1}$) the results from this investigation suggest that the effects of the surface are minimal and thus training on either should have little influence on the levels of impact shock experienced. However any locomotive activities at higher speeds ($5.0 m.s^{-1}$) should consider the surface. Concrete surfaces such as roads and pavements will expose an athlete to larger impact transients than the synthetic sports surface typically used for field hockey. Previous research has identified softer surfaces as reducing the loading rate during running across a population of 6 participants (Dixon *et al.*, 2000). The loading rate has been previously identified earlier in this thesis as being correlated to impact shock. The results from this study report that despite a significant reduction in the loading rate on the softer surface, in one of the participants an increase was reported. The results from the study by Dixon and colleagues demonstrate that similar to the results in this current research, a softer sports surface can reduce detrimental impact loading of the musculoskeletal system. Furthermore, it also reports that individuals react differently to changes in cushioning properties underfoot which is demonstrated in the findings of this current study in terms of footwear and surfaces.

9.4 Conclusion

The Healus Running shoe does not reduce the level of impact shock; however a programme of training in the shoe prior to testing may change this outcome. This should be the basis of further investigation into the effects of this shoe design

The results of this research suggest that participating in high velocity activities on hard concrete surfaces, exposes the musculoskeletal to significantly larger impact shocks. This information can be used by coaches to select suitable activities with the surfaces available to them.

This investigation also identified that during running on synthetic sports surfaces, moulded soccer shoes with hard deep cleats expose the body to larger impact shocks than shoes designed for running and synthetic sports surfaces. Therefore these types of shoes should not be used in field hockey participation on such surfaces. Significant differences in the magnitude of tibial shock between similar footwear designs used by field hockey participants were also identified using the methodologies in this study. The evidence from individual participant provides some assistance for footwear selection. However, the findings of this study demonstrated that shoes can reduce impact shock for one individual while increasing it for another. Therefore, for a participant to be confident of making the correct choice of footwear in relation to exposure to tibial shock, the footwear needs to be tested on an individual basis.

Chapter Ten

Summary

10 Summary

This thesis intended to meet a set of aims and objectives defined in the introductory section (1.2). The first aim was to review the kinematic and kinetic factors influencing overuse injury, and the influence of footwear choice on these factors. The literature review (Chapter 2) identified that field hockey participants employed similar movement strategies to other sports such as soccer and rugby and therefore research from other such sports could help fill the paucity of available information regarding injuries in field hockey. From the literature available, stress fractures in the tibia and feet as well as skin injuries such as corns and calluses, were identified as detrimental occurrences that footwear choice could influence. Ankle sprains were identified and discussed due to the high prevalence of such injuries. However, because the cause of sprains on most occasions is a single traumatic event, they were not considered an overuse injury so were not to be investigated in the scope of this research.

Reducing the likelihood of suffering the types of overuse injuries investigated was identified as being possible through reducing excessive loading causing impact shocks as well as peak pressures applied to the feet. The peak pressures applied to the plantar region were identified as being linked to stress fractures in the foot as well as more commonly to abrasive injuries. A paucity of information regarding the applied pressures to the upper regions of the foot was identified. This formed the basis for the study investigating peak pressures to the sides of the uppers of the feet (Chapter3).

The application of GRFs to the musculoskeletal system and its various links to overuse injuries was investigated. The main conclusions drawn from previous literature reviewed, was that the rate of application of the GRFs and also the direction of application through the footwear were the main factors that could influence injury occurrence during human locomotion. The transient impact shock experienced due to loading of the musculoskeletal system at foot to ground impact was identified as causing many overuse injuries through the musculoskeletal system. The loading rate leading up to the GRF peaks occurring during the initial impact phase was reported to be strongly correlated to the impact shock measured. As well as the rate of loading the direction of the application of the GRF vector was also considered a factor causing injury. This was identified as the angle of the application of the GRF if not along the axial of a bone would apply a bending force. The site of the maximum bending force on a bone had been identified as the site at most risk of injury. This information formed the basis of the study investigating the angular difference between the tibia and the GRF vector during a cutting movement (Chapter 4). While GRF loading rates were reported in the literature to be highly correlated, there were many different methods used to identify loading rates. These included calculating the AVLR, AVL20T80, AVL20T90, AVL50NT50NBW and PVLR. There appeared to be no research comparing such variables. Many studies had used shank mounted accelerometers to directly measure impact shock. Studies collecting accelerometer and force data simultaneously, had identified various correlations between the magnitude of the impact accelerations and the loading rates, calculated in the various ways mentioned. Therefore if GRF variables and impact shock data was going to be used as evidence for the influence of footwear choice on injury, their relationship using the mounting system that would be developed for this thesis needed to be investigated (Chapter 5). By comparing the various methods of identifying peak accelerations through analysis of GRF data, the most suitable variables were identified and used later when comparing footwear (Chapter 8).

Through review of previous research collecting human locomotion data various factors that could influence kinetic and kinematic data were identified. Running speed and strategy, dominant sides, age, gender and bodyweight were all identified as needing to be controlled due to their influence on data. To achieve this, during the footwear testing, the speed of the participants needed to be controlled, the non-dominant side would be chosen to be investigated, and participants would be male with similar ages (18-30) who were healthy and exercised regularly.

The effects of footwear previously investigated; found that the relationship between cushioning and loading of the musculoskeletal system during locomotion was not a simple one. Softer shoes did not necessarily mean better cushioning. The studies identifying this phenomenon concluded that individuals would adjust their movement strategy differently in different shod and surface conditions. These findings identified a need to mechanically test any footwear being investigated in this research to see if this was the case with field hockey footwear (6.2).

The main conclusions drawn from the review of the currently available literature were that footwear had the potential to influence kinetics and kinematics that could reduce the prevalence of overuse injuries in a general population. However, due to the complex nature of human movement, it may be the case that it is not possible to identify a footwear design that will work for all participants.

The next aim was to investigate the pressure distribution between the feet and uppers of footwear; alignments of the tibia with the resultant ground reaction force; and the relationship between ground reaction forces and accelerations in the tibia during human locomotion.

The investigation into localised pressure through the uppers of footwear (Chapter 3), successfully measured peak pressures between footwear and the sides of participants' feet. There is currently no such research available to compare the results to. However, the results found that peak values at the lateral side of the 5th metatarsal during sidestepping (Table 3.1) were about the same magnitude as those reported at the plantar region of the feet during forwards locomotion (Table 7.1). Cutting movements also produced relatively high peak values while starting sprinting and stopping were much lower. This demonstrated that in activities where multidirectional movements were common, footwear designs should consider the distribution of pressure through the uppers to restrict skin injuries common to the lateral side of the 5th metatarsal head.

The relationship between the alignment of the ground reaction force vector and tibia was investigated (Chapter 4). Two different insoles were used to alter the proprioception at the plantar surface of the foot. An increase in the alignment of the tibia and resultant GRF vector was identified at the initial force peak for the insole with a rubber surface for enhanced proprioceptive feedback. The results show that the insole does appear to have an effect on the body's movement compared to the GRF vector. This could have an effect on the occurrence and site of injuries linked to the maximum bending site in the bone. This was a single participant study that identifies an area of further investigation for sports scientists.

Investigating the relationship between GRF characteristics and accelerations along the axial of the tibia (Chapter 5) allowed the most effective GRF characteristics to be identified when considering impact shock measured with the methodology and setup that would be used to measure impact shock later in this thesis (Chapter 9). The accelerometer was attached to the antero-medial distal aspect of the shank (Figure 5.1), which had been the method now widely used in published research. A butterworth low pass filter was used to remove unwanted acceleration signals due to the skin artefact as previously published work had identified. Data from a force plate and accelerometer were collected simultaneously. The data showed that identifying a force peak was not always easily done and for some trials, estimates had to be made. This factor along with the identification of correlations across the population of participants (n=13) identified that calculating the loading rate using PVLR was the most effective as it provided the strongest correlation (r = 0.469) and did not require identification of the impact peak. This was important as the impact peak was not always present and relied on a human to identify each peak. PVLR did not require this and thus should produce more consistent results and allow computer scripts to automatically identify its magnitude as was developed during this research.

The main focus of this thesis was to investigate the effects of footwear choice on various kinetic and kinematic variables measured during locomotion. Firstly, as in previous studies investigating the influence of footwear on kinetics and kinematics, a mechanical test was performed. The mechanical test identified the Saucony Running shoes as the most effective in reducing the rate of loading, followed by the Asics Gel Lethal and then the Gryphon Venom and Gryphon Viper which produced relatively similar results. The Umbro Astroturf followed by the Umbro Moulded provided the least attenuation of the loading.

The influence of the footwear investigated on peak in-shoe pressures during running and jogging did not produce any significant differences over the population of participant (n=8) when comparing footwear. An increase in speed was identified as significant ($P<0.05$) in increasing peak pressures. The usefulness of such a finding could only be really relevant in terms of field hockey, to training situations. By using lower velocity training activities, participants who are prone to suffering an injury linked to peak pressures at the plantar region of the foot can reduce the likelihood of sustaining an injury. The purpose of this thesis was to identify how footwear choice of an individual field hockey participant, could influence the risk of them suffering an overuse injury. To investigate this, data between footwear for individuals was analysed. ANOVAs were run to test for within participant significance levels. For each shoe the number of participants that reported significantly different peak pressures compared to any of the other shoe conditions was reported. This helped identify if a footwear choice could affect the exposure of an individual to peak pressures that could cause an overuse injury. In general this showed that all the shoes could have an effect on injury risk. Further data was reported comparing each of the shoes to each of the other footwear separately for each participant. This method of comparing the footwear identified that the Saucony Running shoes produced the most favourable results amongst the participants investigated. For the shoes designed for synthetic surfaces the Gryphon Viper produced the most favourable results with the least favourable being the Asics Gel Lethal. The Umbro Moulded was clearly the shoe most likely to have a negative effect on peak pressures for an individual. However, what the results clearly show is that for pressure data, a shoe choice that may reduce the risk of injury for one participant may increase the risk for another.

The influence of footwear choice on GRF variables linked to injury was investigated using the same in-shoe pressure measuring system. This allowed more data to be collected, reducing the effects of fatigue and without force plate targeting risks. However the magnitude of the data was lower than would be expected from force plate data. This offset has been identified in previous research and it was reported that the characteristics such as timings of peaks and relative magnitudes were consistent with force data from force plates and could therefore be used. As the magnitude of the data collected was being directly compared to data collected by the same system, when comparing footwear this allowed for fair comparisons of relative magnitudes. Similarly to the peak pressure data and to previous research, higher velocities showed an increase in detrimental loading of the musculoskeletal system. Significantly higher loading rates were identified in the Umbro Soccer Moulded shoes when comparing the means of all the participants. The TPVLR variable also produced significant differences that again identified the Umbro Moulded shoes as experiencing kinetics linked to overuse injury. Analysis of the influence of footwear for each participant was also carried out in the same way as in the study of peak pressures. The results showed that as with the peak pressures footwear could have a positive influence for one participant and a negative influence for another. In general the results comparing individuals influenced by footwear choice did not produce any noticeable favourable footwear amongst the non moulded footwear. The Umbro Moulded were clearly identified as a shoe that should be avoided during locomotion on synthetic sports surfaces.

Impact shock testing using a shank mounted accelerometer reported significant differences between running and jogging and between running on a concrete and synthetic sports surface. Importantly the surfaces were found not to have a significant effect across a population during jogging. This meant that for activities at a pace of $3.3 m.s^{-1}$ or less, a concrete surface would not significantly expose participants to an increased risk of suffering an overuse injury. Furthermore such low velocity activities produced much smaller impact accelerations. The increase from $3.3 m.s^{-1}$ to $5.0 m.s^{-1}$ in many cases more than doubled the magnitude of the acceleration. Therefore it was reported that running should be the area of concern for overuse injuries. Furthermore, during such activities, using a synthetic sports surface compared to concrete surfaces can assist in significantly reducing the likelihood of such injuries.

The footwear tested for accelerations at the tibia included the Healus footwear (Figure 9.1). Comparing the effects of footwear on the means of participants reported similar results to the GRF data collected. Once more the Umbro Moulded produced significantly detrimental kinetics. The peak accelerations measured at the tibia were found to be significantly higher in this footwear during running on the synthetic sports surface. The Umbro Moulded footwear could not be tested on concrete due to safety concerns. During jogging, no significant differences in footwear conditions were reported. Comparing the effects of footwear choice on the individuals in the same manner for the GRF and peak pressure research, identified the Umbro Moulded as increasing the impact shock magnitude in most participants compared to all the other footwear individually. The other shoes produce positive and negative results suggesting that the effects of similar types of footwear on impact shock are very dependent on the individual. The Healus® footwear in general compared to the other footwear (excluding the Umbro Moulded) did not perform favourably. However from the discussion of previous research it was concluded that extensive training in such footwear could provide different results.

This thesis investigated the influence of various types of footwear on kinetics linked to overuse injuries. In a general population the moulded footwear was found to expose the athletes to detrimental loading of the musculoskeletal system. Amongst the other footwear which consisted of running shoes, soccer synthetic surface shoes and hockey synthetic surface shoes, it was found that significant differences were only found within individuals. A summation of all the results comparing each of the footwear to each of the other footwear can be found in the appendix (Table 12.1 to 12.7). A summation of all the results comparing each shoe to all of the other shoes can also be found in the appendix (Table 12.8). For these two methods of analysing the results of footwear, the method identifying differences between footwear with each of the other footwear is more valuable. This method allows a field hockey participant to select the footwear that is most likely to reduce their risk of being exposed to excessive kinetics. However this is a relatively small sample of participants so is not conclusive. More important than this is that while footwear can show general trends, it is clear that footwear that may protect most participants from detrimental kinetics may expose an individual to an increased risk. Therefore the findings of this thesis conclude that individual participants especially those from a high risk of injury group, should be individually assessed in the various footwear.

Using the computer program developed in Matlab within this thesis will provide a quick and practical assessment for field hockey participants. This will allow for an informed choice of footwear and its influence on overuse injuries.

Based on the outcome of this study the following protocol is suggested for any future assessment of footwear choice within any sport.

1. *The surface characteristics of the data collection area should match the surface for which the individual will be typically participating in the activity.*
    - *The other parameters of the data capture area should follow normally accepted procedures including appropriate placement of timing gates.*
2. *The methodology for data capture should be standardised to provide valid and reliable data.*

- *Attach the accelerometer tightly to the skin covering the lower aspect of the tibia.*
- *Insert a pressure sensor in the shoe, and ask the participant to put the shoe on carefully checking the senor has not moved.*
- *Allow the participant appropriate time to become accustomed to the shoe.*
- *Ideally a wireless pressure and accelerometer system should be employed. If these are not readily available care should be taken to achieve a more natural movement.*

3. *Accepted and previously validated procedures for data collection should be followed, such as:*
    - *Record an appropriate number of trials which would provide a suitable amount of data for statistical analysis.*
    - *Where appropriate randomise the order of the shoes or conditions being assessed.*
4. *Results should be reported to provide the relevant information that can be used by a coach or sports scientist to make an informed choice as to the most suitable footwear.*

10.1 Summary of the contribution of this thesis to scientific literature

This section identifies the main contribution this thesis has made to scientific literature.

This research identified that:

- *The magnitude of peak pressures applied to the uppers of the foot can be similar to those experienced in the plantar region of the foot which cause overuse injuries.*
- *Footwear insoles can influence the alignment of the tibia and the resultant GRF vector.*
- *Increased cushioning in footwear does not necessarily protect the body from detrimental loading during field hockey participation.*
- *From the designs investigated, moulded soccer shoes expose field hockey participants to an increase risk of experiencing an overuse injury compared to shoes made for synthetic sports surfaces and running.*
- *And that high velocity activities in the footwear tested on concrete significantly increases the magnitude of transient impact shocks experienced when compared to the same footwear on a synthetic sports surface.*

Chapter Eleven

Directions of Further Study

11 Directions of Further Study

This section identifies where findings from this thesis suggest further research advancing scientific knowledge. Furthermore, this section also identifies where further research conducted in this thesis required further investigation to increase scientific understanding.

- There is currently a paucity of research investigating individual's movements during field hockey matches and training, and their relation to overuse injury occurrence. Simply recording the duration and frequency of activities is not sufficient, as the intensity and types of physical activities will be a factor that can influence injury occurrence. Therefore there is a need for further research to be carried out thorough study of hockey participation, recording the types of activity, their duration and frequency. Such a study may identify specific training strategies including periods of rest, that may help coaches design schedules of participation that help reduce the occurrence of injuries.

- The alignment of the tibia and the GRF vector may be an area of concern, however little is known of the influence of this variable on injury occurrence. Future research comparing patients with a history of injuries in the tibia could be compared to a population with no such injury history. If significant differences between the two populations were found it may assist in identifying favourable adjustments in kinetic and kinematic data to reduce the risk of injury.

- Research in this thesis investigating peak pressures, identified that peak pressures similar to those experienced in the plantar region of the foot were present between the uppers of the footwear and the foot. Further investigation into the distribution of pressure through the uppers of footwear may assist in footwear design. By testing the influence of footwear on sites on the foot that are known to be of concern, it may be possible to identify suitable footwear designs that reduce the occurrences of peak pressures.

- As well as directly measuring applied forces to the musculoskeletal system, it is possible to measure the deformation of bone tissue directly. An investigation using a surgical staple and strain gauge system found peak tibia deformation occurs between 20-42ms after ground contact and was up to eight times higher than when the participants were stood still on a single leg (Rolf *et al.*, 1997). Bone deformations measured by the same group of researchers reported that peak mean deformations of the tibia were localised with different sites being exposed to smaller or larger strains depending on the movements being performed (Ekenman *et al.*, 1998). This data provides valuable evidence that specific locomotive movements place large amounts of stress at specific sites of the bone and therefore adjusting the movement strategy or reducing the magnitude of the loading can assist in reducing the localised stresses applied repeatedly to certain sites in the musculoskeletal system. Research into the effects of loading on bones at specific locations while wearing field hockey specific footwear could provide further information that may assist in footwear choice.

- This study collected different kinetic data on the same footwear at different times. Collecting data such as in-shoe pressure and tibial accelerations simultaneously, may provide further understanding of the effects of field hockey specific footwear on factors relating to injury. However, the restrictions of the equipment and environment may provide un-natural movement strategies and must be carefully considered.

- This research has concentrated on repeated locomotion trials. By collecting data during simulated or actual match situations, more realistic data may be obtained. The main problem to overcome for such a study would be the danger of damaging the equipment and the participant. As biomechanical systems become smaller and lighter and by using data loggers or telemetric systems, in the future this should be possible.

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Appendix

12 Appendix

Table 12.1 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Asics Gel Lethal shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus Running	
Value of the Asics Gel Lethal Variable Below, Compared to the Condition Above		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	2	2	1	6	0	1	0	0	2		
	Running on ISSS	0	2	1	2	4	2	2	2	0	2		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	0	5	0	2	0	4	0	1	0	6		
	Running on ISSS	0	4	1	2	1	0	1	1	0	3		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	0	2	0	1	3	1	1	1	1	2		
	Running on ISSS	0	2	0	1	2	0	0	0	1	2		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	2		
	Running on ISSS	0	3	0	1	4	0	0	1	0	3		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	4	1	0	3	0	0	0	0	5		
	Running on ISSS	0	1	0	1	0	2	0	1	0	2		
TVFP1	Jogging on ISSS	1	0	0	0	0	1	2	0	1	0		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
TPVL	Jogging on ISSS	0	0	0	0	0	0	1	0	4	0		
	Running on ISSS	0	0	2	0	0	0	0	0	3	0		
VFP1	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
AVLR	Jogging on ISSS	0	0	0	0	5	0	0	0	0	0		
	Running on ISSS	0	0	0	0	2	0	0	0	0	0		
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	2	0	0	0	0	0		
	Running on ISSS	0	0	0	0	2	0	0	0	0	0		
AVL20T90	Jogging on ISSS	1	0	0	0	3	0	0	0	1	0		
	Running on ISSS	0	0	0	0	3	0	0	0	0	0		
AVL20T80	Jogging on ISSS	1	0	0	0	4	0	0	0	0	0		
	Running on ISSS	0	0	0	0	3	0	0	0	0	0		
PVLR	Jogging on ISSS	0	0	0	0	5	0	0	0	0	0		
	Running on ISSS	0	0	4	0	3	0	0	0	0	0		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	0	6	0	1	0	0	0	2	0
	Jogging on Concrete	0	0	0	1			0	1	0	2	0	1
	Running on OSSS	0	2	1	2	4	0	1	1	0	0	1	0
	Running on Concrete	0	0	0	0			0	0	0	0	0	1

Table12.2 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Saucony Running shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Asics Gel Lethal		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus Running	
Value of the Saucony Running Variable Below Compared to the Condition Above:		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	2	0	2	0	7	0	1	0	0	0		
	Running on ISSS	2	0	2	1	7	0	1	1	1	0		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	5	0	2	0	2	0	3	0	0	1		
	Running on ISSS	4	0	1	0	2	0	2	0	1	0		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	2	0	1	0	5	1	1	0	2	0		
	Running on ISSS	2	0	1	0	5	0	1	0	2	0		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0		
	Running on ISSS	1	0	0	0	0	1	0	0	0	1		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	4	0	2	0	5	0	1	0	0	1		
	Running on ISSS	3	0	2	0	6	0	1	0	0	1		
TVFP1	Jogging on ISSS	0	1	0	1	0	4	2	0	1	0		
	Running on ISSS	0	0	0	1	0	3	0	0	0	0		
TPVL	Jogging on ISSS	0	0	0	0	0	2	0	0	0	0		
	Running on ISSS	0	0	0	0	0	2	0	0	0	0		
VFP1	Jogging on ISSS	0	0	0	1	0	2	0	0	0	1		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
AVLR	Jogging on ISSS	0	0	0	0	3	0	0	0	0	0		
	Running on ISSS	0	0	0	0	3	0	0	0	0	0		
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	3	0	0	0	0	0		
	Running on ISSS	0	0	0	0	2	0	0	0	0	0		
AVL20T90	Jogging on ISSS	0	1	0	2	4	0	0	1	0	0		
	Running on ISSS	0	0	0	0	1	0	0	0	0	0		
AVL20T80	Jogging on ISSS	0	1	0	2	3	0	0	2	0	0		
	Running on ISSS	0	0	0	0	5	0	0	0	0	0		
PVLR	Jogging on ISSS	0	0	1	0	5	0	0	0	0	0		
	Running on ISSS	0	0	2	0	4	0	0	0	0	0		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	1	3	0	1	1	0	1	2	0
	Jogging on Concrete	0	0	0	0			0	1	0	1	2	1
	Running on OSSS	2	0	0	1	5	0	0	0	0	0	2	0
	Running on Concrete	0	0	0	0			1	0	1	0	1	0

Table12.3 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Umbro Astroturf shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Asics Gel Lethal		Saucony Running		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus Running	
Value of the Umbro Astroturf Variable Below, Compared to the Condition above :		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	1	2	0	2	6	0	1	2	0	1		
	Running on ISSS	2	1	2	1	6	0	1	0	0	1		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	2	0	0	2	2	1	2	0	0	2		
	Running on ISSS	2	1	0	1	1	0	1	0	0	1		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	0	0	1	5	0	0	1	1	1		
	Running on ISSS	1	0	0	1	4	0	1	0	2	0		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	1	0	0	1	2		
	Running on ISSS	1	0	0	0	0	1	1	2	0	1		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	1	0	2	4	0	0	1	0	3		
	Running on ISSS	1	0	0	2	6	0	1	1	0	1		
TVFP1	Jogging on ISSS	0	0	1	0	0	3	2	0	1	0		
	Running on ISSS	0	0	1	0	2	0	0	0	0	0		
TPVL	Jogging on ISSS	0	0	0	0	0	3	0	0	0	0		
	Running on ISSS	0	2	0	0	0	3	0	0	0	0		
VFP1	Jogging on ISSS	0	0	1	0	0	1	0	0	0	0		
	Running on ISSS	0	0	0	0	0	2	0	1	0	0		
AVLR	Jogging on ISSS	0	0	0	0	0	1	0	1	0	1		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	2	1	0	1	0	1		
	Running on ISSS	0	0	1	0	2	0	0	0	0	0		
AVL20T90	Jogging on ISSS	0	0	2	0	0	0	0	0	0	1		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
AVL20T80	Jogging on ISSS	0	0	2	0	3	1	0	0	0	1		
	Running on ISSS	0	0	0	0	3	0	0	0	0	0		
PVLR	Jogging on ISSS	0	0	0	1	2	1	0	0	0	0		
	Running on ISSS	0	0	0	2	0	0	0	1	0	0		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	1	1	1	5	0	0	1	0	1	1	0
	Jogging on Concrete	1	0	0	0			0	1	0	1	1	1
	Running on OSSS	2	1	1	0	5	0	1	1	2	0	3	0
	Running on Concrete	0	0	0	0			0	1	0	0	0	1

Table 12.4 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Umbro Moulded shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Gryphon Venom		Gryphon Viper		Healus Running	
Value of the Umbro Moulded Variable below, Compared to the Condition above:		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	6	0	7	0	6	1	4	0	7		
	Running on ISSS	0	6	0	7	0	6	0	6	0	6		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	4	0	0	2	1	2	1	1	0	3		
	Running on ISSS	0	1	0	2	0	1	0	2	0	2		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	3	1	5	0	5	0	4	0	5		
	Running on ISSS	0	2	0	5	0	4	0	2	0	4		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	1	0	0	0	0	0		
	Running on ISSS	2	0	1	0	1	0	1	0	1	1		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	3	0	5	0	4	0	4	0	7		
	Running on ISSS	0	4	0	6	0	6	0	4	0	7		
TVFP1	Jogging on ISSS	1	0	4	0	2	0	6	0	6	0		
	Running on ISSS	0	0	3	0	2	0	1	0	2	0		
TPVL	Jogging on ISSS	0	0	2	0	3	0	3	0	5	0		
	Running on ISSS	1	0	2	0	3	0	1	0	3	0		
VFP1	Jogging on ISSS	0	0	2	0	1	0	0	0	0	0		
	Running on ISSS	0	0	0	0	2	0	0	0	0	0		
AVLR	Jogging on ISSS	0	5	0	3	1	0	0	6	0	4		
	Running on ISSS	0	2	0	3	0	0	0	3	0	3		
AVL50NT50NBW	Jogging on ISSS	0	2	0	3	0	2	0	4	0	5		
	Running on ISSS	0	2	0	2	0	2	0	2	0	2		
AVL20T90	Jogging on ISSS	0	3	2	2	0	0	0	4	0	5		
	Running on ISSS	0	2	0	0	0	0	0	2	0	2		
AVL20T80	Jogging on ISSS	0	4	0	3	1	3	0	4	0	5		
	Running on ISSS	0	4	0	5	0	3	0	4	0	4		
PVLR	Jogging on ISSS	0	0	0	1	1	2	0	0	0	0		
	Running on ISSS	0	3	0	4	0	0	0	3	0	2		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	**0**	**6**	0	3	**0**	**5**	**0**	**6**	**0**	**6**	1	4
	Jogging on Concrete												
	Running on OSSS	**0**	**4**	**0**	**5**	**0**	**5**	**0**	**7**	**0**	**5**	**0**	**6**
	Running on Concrete												

Table12.5 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Gryphon Venom shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Viper		Healus Running	
Value of the Gryphon Venom Variable Below, Compared to the Condition above:		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	0	1	0	1	1	2	4	1	0	3		
	Running on ISSS	2	2	1	1	0	1	6	0	0	0		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	1	0	0	3	0	2	1	1	0	3		
	Running on ISSS	1	1	2	0	1	0	2	0	0	1		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	1	1	0	1	1	0	4	0	0	0		
	Running on ISSS	0	0	0	1	0	1	2	0	1	0		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0		
	Running on ISSS	0	0	0	0	2	1	0	1	0	0		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	0	0	0	1	1	0	4	0	0	2		
	Running on ISSS	1	0	0	1	1	1	4	0	0	1		
TVFP1	Jogging on ISSS	0	2	0	2	0	2	0	6	0	1		
	Running on ISSS	0	0	0	0	0	0	0	1	0	0		
TPVL	Jogging on ISSS	0	1	0	0	0	0	0	3	0	0		
	Running on ISSS	0	0	0	0	0	0	0	1	1	0		
VFP1	Jogging on ISSS	0	0	0	0	0	0	0	0	0	0		
	Running on ISSS	0	0	0	0	1	0	0	0	0	0		
AVLR	Jogging on ISSS	0	0	1	0	6	0	0	0	0	0		
	Running on ISSS	0	0	0	0	0	0	3	0	0	0		
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	1	0	4	0	0	0		
	Running on ISSS	0	0	0	0	0	0	2	0	0	0		
AVL20T90	Jogging on ISSS	0	0	2	0	0	0	4	0	1	0		
	Running on ISSS	0	0	0	0	0	0	2	0	0	0		
AVL20T80	Jogging on ISSS	0	0	2	0	0	0	4	0	1	0		
	Running on ISSS	0	0	0	0	0	0	4	0	0	0		
PVLR	Jogging on ISSS	0	0	0	0	0	0	6	0	0	0		
	Running on ISSS	0	0	0	0	0	0	3	0	0	0		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	1	1	1	1	0	7	0	0	2	2	1
	Jogging on Concrete	1	0	1	0	1	0			0	0	2	0
	Running on OSSS	1	1	0	0	1	1	7	0	1	0	3	1
	Running on Concrete	0	0	0	1	1	0			0	0	0	1

Table 12.6 Number of Participants reporting significant (P<0.05) Larger (L) and Smaller (S) GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Gryphon Viper shoes to the other footwear investigated. Red = Larger number of positive differences relating to injury. Yellow = Smaller number of positive differences relating to injury.

		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Healus Running	
Value of the Gryphon Viper Variable Below, Compared to the Condition above :		S	L	S	L	S	L	S	L	S	L	S	L
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	2	0	0	0	1	0	7	0	3	0		
	Running on ISSS	2	0	0	1	1	0	6	0	0	0		
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	6	0	1	0	2	0	3	0	3	0		
	Running on ISSS	3	0	0	1	1	0	2	0	1	0		
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	2	1	0	2	1	1	5	0	0	0		
	Running on ISSS	2	1	0	2	0	2	4	0	0	1		
Peak Pressure Under the Calcaneum	Jogging on ISSS	0	0	0	0	1	0	0	0	0	0		
	Running on ISSS	2	0	1	0	1	0	1	1	0	0		
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	5	0	1	0	3	0	7	0	3	0		
	Running on ISSS	3	0	1	0	1	0	7	0	1	0		
TVFP1	Jogging on ISSS	0	1	0	1	0	1	0	6	1	0		
	Running on ISSS	0	0	0	0	0	0	0	2	0	0		
TPVL	Jogging on ISSS	0	4	0	0	0	0	0	5	0	0		
	Running on ISSS	0	3	0	0	0	0	0	3	0	1		
VFP1	Jogging on ISSS	0	0	1	0	0	0	0	0	0	0		
	Running on ISSS	0	0	0	0	0	0	0	0	0	0		
AVLR	Jogging on ISSS	0	0	0	0	1	0	4	0	0	0		
	Running on ISSS	0	0	0	0	0	0	3	0	0	0		
AVL50NT50NBW	Jogging on ISSS	0	0	0	0	0	0	5	0	0	0		
	Running on ISSS	0	0	0	0	0	0	3	0	0	0		
AVL20T90	Jogging on ISSS	0	1	0	0	1	0	5	0	0	1		
	Running on ISSS	0	0	0	0	0	0	2	0	0	0		
AVL20T80	Jogging on ISSS	0	0	0	0	1	0	5	0	1	0		
	Running on ISSS	0	0	0	0	0	0	4	0	0	0		
PVLR	Jogging on ISSS	0	0	0	0	0	0	7	0	0	0		
	Running on ISSS	0	0	0	0	0	0	2	0	0	0		
Tibial Axial Acceleration Magnitude	Jogging on OSSS	0	0	1	0	1	0	6	0	2	0	2	1
	Jogging on Concrete	2	0	1	0	0	0			0	0	1	0
	Running on OSSS	0	0	0	0	0	2	5	0	0	1	2	0
	Running on Concrete	0	0	0	1	0	0			0	0	0	1

Table12.7 Number of Participants reporting significant (P<0.05) positive and negative GRF, peak pressure and tibial acceleration, for characteristics linked to overuse injuries. Comparing Healus shoes to the other footwear investigated

Healus Running Compared to:	Asics Gel Lethal		Saucony Running		Umbro Synthetic Turf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus Running	
Relative Tibial Axial Acceleration Magnitude	L	H	L	H	L	H	L	H	L	H	L	H	L	H
Jogging on Synthetic Surface	0	2	0	2	0	1	4	1	1	2	1	2		
Running on Synthetic Surface	1	0	1	2	1	0	**6**	**0**	0	2	0	1		
Jogging on Concrete	0	1	0	2	0	3	1	3	0	2				
Running on Concrete	1	0	0	1	1	0	1	0	1	0				

Key:

Yellow = Greater amount of reported higher mean peak axial accelerations.

Red = Greater amount of reported lower mean peak axial accelerations.

Black = Result not available.

L=Lower

H=Higher

Table 12.8 Number of participants reporting significant differences in the magnitude of kinetic variables when comparing the effects of each footwear condition to all other pairs. Red = Positive effect, Yellow = negative effect

Footwear Condition		Asics Gel Lethal		Saucony Running		Umbro Astroturf		Umbro Moulded		Gryphon Venom		Gryphon Viper		Healus Running	
Relationship to Injury		+	-	+	-	+	-	+	-	+	-	+	-	+	-
Peak Pressure Under the 1st Metatarsal Head	Jogging on ISSS	6	2	7	0	6	3	1	7	4	3	7	0	-	-
	Running on ISSS	7	2	7	1	8	3	0	7	7	2	6	1	-	-
Peak Pressure Under the 2nd and 3rd Metatarsal Heads	Jogging on ISSS	0	6	6	1	3	2	4	3	1	5	6	0	-	-
	Running on ISSS	6	3	0	6	1	5	1	0	2	3	2	4	-	-
Peak Pressure Under the 4th and 5th Metatarsal Heads	Jogging on ISSS	3	3	7	1	5	1	2	5	5	1	5	2	-	-
	Running on ISSS	3	4	6	0	5	1	0	6	3	1	4	2	-	-
9Peak Pressure Under the Calcaneum	Jogging on ISSS	0	1	0	0	1	2	0	1	0	0	3	1	-	-
	Running on ISSS	1	2	2	1	2	2	2	1	2	1	2	1	-	-
Peak Pressure Under the Entire Planter Region	Jogging on ISSS	3	7	6	1	4	3	0	7	4	3	8	0	-	-
	Running on ISSS	3	4	7	1	6	2	0	7	7	1	8	0	-	-
TVFP1	Jogging on ISSS	0	0	0	2	1	0	2	0	0	0	1	0	-	-
	Running on ISSS	0	0	0	0	0	2	2	0	1	0	0	0	-	-
TVFP2	Jogging on ISSS	1	3	4	2	3	3	0	7	7	0	6	1	-	-
	Running on ISSS	0	1	3	0	2	0	0	3	0	0	0	0	-	-
TPVL	Jogging on ISSS	0	4	3	0	3	0	0	6	4	0	5	0	-	-
	Running on ISSS	1	2	2	0	3	0	0	3	1	1	3	0	-	-
AVLR	Jogging on ISSS	4	0	2	1	0	3	1	6	6	0	3	0	-	-
	Running on ISSS	2	0	3	0	0	0	0	3	3	0	3	0	-	-
AVL50NT50NBW	Jogging on ISSS	1	0	2	0	1	1	1	5	4	0	6	0	-	-
	Running on ISSS	2	0	2	1	3	0	0	3	2	0	3	0	-	-
AVL20T90	Jogging on ISSS	3	0	2	3	4	0	3	4	5	0	4	1	-	-
	Running on ISSS	3	0	1	0	0	0	0	3	2	0	2	0	-	-
AVL20T80	Jogging on ISSS	4	0	2	3	6	1	1	4	5	0	4	1	-	-
	Running on ISSS	4	0	5	0	3	0	0	5	4	0	4	0	-	-
PVLR	Jogging on ISSS	5	0	5	0	3	1	1	7	6	0	6	0	-	-
	Running on ISSS	3	0	4	0	0	2	0	4	3	0	2	0	-	-
Tibial Axial Acceleration Magnitude	Jogging on OSSS	6	0	5	1	5	1	1	6	6	2	7	1	4	2
	Jogging on Concrete	2	3	3	1	4	2	-	-	3	2	2	3	1	6
	Running on OSSS	4	2	5	1	6	1	0	7	7	0	6	0	6	3