The Effect of Playing Surfaces and Footwear on the Biomechanical Response of Soccer Players

Submitted by Daniel Craig Low to the University of Exeter as a thesis for the degree of Doctor of Philosophy in Sports and Exercise Science, January 2010.

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Abstract

This thesis describes three studies investigating biomechanical responses to changes in shoe-surface combinations in soccer. In the first study, six male participants (21.7 [S.D. 2.2] yrs, 74.0 [S.D. 6.9] kg [March], 74.6 [S.D. 6.9] kg (May), footwear size 10 -11) performed running and turning movements on natural and third generation artificial surfaces whilst wearing different soccer specific footwear. This was performed at two times of the year where contrasting weather conditions were experienced. It was observed that there were significant differences when the natural and third generation artificial turf surfaces were compared. These differences however, were dependent on the type of movement, time of year and biomechanical measurement used. Each surface was also compared between the two test occasions. The main finding was that for both running and turning peak pressures and peak pressure loading rates were significantly greater in May (when the surfaces were mechanically hard) compared with the same surface in March. It was concluded that comparisons of third generation surfaces with natural turf are dependent on the specific properties of the surfaces and cannot be generalized for all such surfaces.

A critical design feature of third generation surfaces that will influence biomechanical comparisons with other playing surfaces is the shock pad layer. In the second experimental chapter, ten male participants (20.9 yrs [S.D. 2.5], 83.2 kg [S.D. 7.1], footwear size 10 -11) were used to assess the effect of two different shock pad densities (55g and 65g) (Arpro® Expanded polypropylene BF2455W, 24mm S.D. 0.5mm thick, Brock International) on the lower extremity loading. These participants were also used to assess the biomechanical adaptations that occur with the inclusion of a 10 mm Sorbothane® heel insert or a Sorbothane® cushioning insole (Sorbo products division, Lancashire, UK), which have been associated with reducing overuse injury including that to the Achilles tendon. The footwear was also assessed for the risk of sustaining lateral ankle ligament damage. It was shown that whilst turning, peak impact force (taken using in-shoe pressure system) was significantly lower on the more cushioned shock pad as was peak pressure at the first metatarsal. Likewise, the time to peak impact force was significantly longer with the heel insert. However, despite the association between the heel insert and reduced dorsi-flexion, no significant differences were observed for this measurement between the footwear conditions. Peak plantar flexion was significantly greater with the heel insert whilst turning

suggesting an increased loading of the lateral ankle ligaments, although rearfoot inversion was not significantly different. This study demonstrated the potential role of shock pad cushioning in providing protection from impact related injury in soccer, whilst cushioning inserts were not found to provide a protective effect. For heel inserts, the possibility of a negative influence on rearfoot stability was highlighted. It was suggested that the estimation of internal loads may reveal more regarding the specific role of cushioning interfaces and heel inserts in protecting from injury.

In the final research chapter, nine male soccer players (83.4 kg [S.D. 5.8], 23 yrs [S.D. 3.7]) performed running and turning movements for the same conditions described in study two. The peak plantar flexion moment, Achilles tendon force and average loading rate of these measurements, were used to assess Achilles tendon loading. Likewise, peak dorsi-flexion and eversion moments were collected to assess the lateral ankle loading. Group analysis did not reveal any significant differences in these variables. Individual data showed that the response to heel insert intervention was specific to the participant. Some participants exhibited a reduced Achilles tendon force or average loading rate, suggesting a reduced risk of injury with the heel insert. However, it was observed that eversion and dorsi-flexion moment and average loading rates increased in some participants, suggesting that these participants were at an increased risk of lateral ankle ligament injury with the heel insert. Likewise, one participant experienced significantly greater peak Achilles tendon force, also indicating a greater risk of injury to this structure.

The overall conclusions gained from these studies are that the design of the footwear and playing surfaces are worth considering in the quest to reduce injury risk. It was also highlighted that the choice of shock pad density for a third generation artificial surface can be influential in the protection of the athlete even when the surface is new, particularly when turning. Finally, although the use of heel inserts has proven successful in the reduction the Achilles tendon injury, the lack of significant differences for group comparisons suggests that the mechanisms behind the success is still unclear. However, although heel inserts may prove useful in the reduction of Achilles tendon injury, the observation that significant increases in the measurements associated with acute ankle ligament damage and chronic Achilles tendon injury, suggests that heel lift may not be suitable for some individuals.

Acknowledgements

Whilst collecting the data for the studies presented in this thesis, I have relied on many of participants. As such I would like to thank them for sparing their valuable time to take part in these investigations.

I would also like to thank my supervisor Dr Sharon Dixon for the academic support and encouragement provided throughout which were much needed and appreciated. I would also like to give special thanks to Dr Alan Barker for the valued ideas and advice provided during the writing up process. Also, I would like to express my appreciation for the general support and advice provided by Dr Chris Mills, Dr Vicky Stiles, Professor Andrew Sparkes and Mr Michael Nunns.

Lastly and most importantly I would like to thank my family; my mum and dad, brothers, Mathew and Stephen and sister Jessica, for supporting me when times became difficult, and to my wife Louise, whose unwavering support and advice has got me through the toughest of periods and who without, I may not have succeeded.

Contents

Definition of abbreviations and symbols

Abbreviations

3g: Third generation artificial turf surface

COM: Centre of Mass

COP: Centre of pressure

Deg: Degree

DLT: Direct Linear Transformation

GRF: Ground Reaction Force

S.D: Standard deviation

VGRF: Vertical Ground Reaction Force

Measurement units/symbols

BW (Body weights): Magnitude of force for an individual normalized by their body weight, i.e., peak impact force/body weight.

N (Newtons): Unit of force ($1N = 1Kg.m.s^{-2}$)

N/cm^2 (Newton per centimetre square): Measurement of pressure

$N/cm^2.ms$ (Newton per centimetre square per millisecond): Measurement of pressure loading rate

N.m (Newton meters): Measurement of moments/torques

N/ms (Newtons per milli-second): Measurement of loading rate

Ms (Millisecond): Measurement of time

N/sec (Newtons per second): Measurement of loading rate

Pascal : Measurement of pressure

Sec (Second): Measurement of time

° (Degrees): Measurement of a angle

®: Registered trade mark symbol

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Average loading rate

Peak force or moment/occurrence time Equation 5.3

Direct Linear Transformation

$$x + \delta x = \frac{L1x+L2y+L3z+L4}{L5x+L6y+L11z+1} \qquad \text{Equation 2.1}$$

$$y + \delta y = \frac{L5x+L6y+L7z+L8}{L9x+L10y+L11z+1} \qquad \text{Equation 2.2}$$

Instantaneous loading rate

$L_1 = (x_i+1 - x_i-1) / 2\Delta t$ Equation 3.1

L_1= instantaneous loading

x_i = vertical force for the ith field

Δt = time interval

Ankle and forefoot joint centres

Ankle joint centre = L + ((0.5 * A) + R)* X Equation 5.1

Forefoot centre = F + ((0.5* W) + R) * X Equation 5.2

Where

L= location of the lateral maleolus marker

A= ankle width

R= Radius of marker

X= X foot (see appendix)

F= location of the Fifth metatarsal marker

W = forefoot width

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Chapter 4

Conferences

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1. Introduction

Soccer is one of the most participated sports in the world, with approximately 200,000 professional, and 240 million amateur, participants (Judge & Dvorak, 2004). Involvement in soccer can contribute towards an active lifestyle, which has been shown to impact self- esteem and self-confidence and reduce your risk of developing a serious illness (NHS direct, 2009). However, those who participate in soccer are also at a considerably greater risk of injury compared to those who participate in other sports such as handball, basketball, hockey, gymnastics, cycling, boxing, judo, rowing, swimming, and mountain hiking (Junge, Dvorak, Graf-Baumann & Peterson, 2004; Pons-Villanueva, Seguí-Gómez, & Martínez-González 2009; Weightman & Browne, 1975; Yde & Nielsen, 1990). The occurrence of injury is a fundamental problem which can influence the psychological, physiological and biomechanical components of performance, as well as the emotional state of the participant away from the sporting arena. The experience of musculoskeletal injury can also be financially taxing on both the sports clubs and the sports player. Likewise, sports participation significantly contributes to the number of a musculoskeletal injuries treated by the National Health Service, which costs approximately £590 million per year (Cullen & Batt, 2005). Further still, the occurrence of a minor injury in the first instance can put a participant at an increased risk of a more severe re-injury, accentuating the effect of the injury further (Bahr & Holme, 2003). The contribution of these factors highlights the need to reduce the risk of injury in soccer. Therefore, the general aim of this thesis is to use biomechanical techniques to improve our understanding of the biomechanical risk factors associated with injury in soccer and how they can be changed to affect the performer. Conducting such research may enable recommendations to be developed on how future injuries could be avoided.

To achieve such an aim, it is first important to identify the extent of the injury problem in soccer. This is described in chapter 2, along with the most common injuries and the mechanism behind the injury occurrence. Evidence is also presented that shows a preseason and early season injury bias in soccer, where injury rates are higher at this time than during others (Ekstrand & Gillquist, 1983a; Hawkins, Hulse, Wilkinson, Hodson, & Gibson 2001; Orchard, 2002; Waldén, Hagglund, & Ekstrand, 2005a; Woods, Hawkins, Hulse, & Hodson, 2002; Woods, Hawkins, Hulse, & Hodson 2003). These periods coincide with the summer months. During this time there is low rain fall and higher temperatures (Woods et al., 2002; Meyers & Barnhill 2004), which may explain, in part, Woods and colleagues (2002) observation that the playing surface was hard and dry when the majority of injuries occurred during preseason. This may have influenced the cushioning provided by the natural turf surfaces and the surface traction characteristics, both of which are linked to the risk of injury at this time (Woods et al., 2002; 2003; Orchard, 2002). These factors are also commonly associated with injury in other sports such as American football, rugby and Australian rules football (Andresen, Hoffman, & Barton, 1989; Lee & Garraway, 2000; Orchard & Powell, 2003).

The magnitude of traction is also influenced by the footwear that is worn (Chomiak, Junge, Peterson & Dvorak, 2000; Lees & Nolan, 1998). In soccer, players commonly wear boots with studs (screw-in and moulded) or soccer trainers with pimples. The height and number of the studs or pimples determines the depth of penetration into the ground and interact with the surface to establish the level of traction that is provided (Santos, Carline, Flynn, Pitman, Feeney, Patterson & Westland, 2001). When the traction becomes too great, the stability of the ankle joint can be compromised, causing the risk of acute injury to increase (Blazevich, 2004). The studs are also thought to determine the loading on specific structures such as the metatarsal and heel, leading to possible overuse injury of these structures (Shorten, 1998; Coyles & Lake, 1999), particularly when performing on hard surfaces (Queen, Charnock, Garrett Jr, Hardaker, Sims, & Moorman III, 2008).

Hard playing surface conditions and inappropriate choice of footwear have been associated with many injuries. In particular, they are found to be related to Achilles tendon, metatarsal and anterior talofibular ligament (ATFL) injury. To understand how the playing surface and footwear changes the participant's risk of injury, biomechanical and mechanical measurements are commonly taken. Changes in these measurements are associated with a change in injury risk. Thus, the observation that the biomechanical characteristics of a soccer player had been adapted when a risk factor had been altered would indicate a change in the injury risk. However, the observation of biomechanical changes to risk factors can depend on the movement patterns performed during testing. In chapter 2, an overview of existing biomechanical and mechanical literature investigating the effect of these risk factors is provided. A review into the strategies in which risk factors can be modified is also provided.

Although some of the currently published footwear-surface research is presented in chapter 2, much is not soccer specific and thus the understanding of surface-footwear interactions in soccer is somewhat limited. This is because research into the effect of footwear interactions on natural turf surfaces is hampered by logistical complications of placing natural turf into a biomechanics laboratory (Stiles, Dixon, Guisasola, & James, 2008). Likewise, it is difficult to test natural turf outside due to problems with standardising the surfaces and because of the inflexibility of some biomechanical equipment. In the studies that have compared changes in natural turf, soil density has been manipulated in a soil bin with no moisture or grass cover (Dixon, James, Blackburn, Pettican, & Low, 2008) or different natural soil-turf types have been imported into a laboratory in trays, where grass growth is time consuming and may be difficult to maintain (Stiles et al., 2008). In both instances the natural turf tested may not be replicable of natural turf found on a soccer pitch.

Artificial surfaces have been proposed as a method by which sports-related lower limb injury may be reduced. Third generation artificial surfaces are designed specifically for use in soccer. However, similar problems exist when trying to biomechanically compare natural and third generation turf surfaces. Therefore, there is also a limited number of these studies (Ford, Manson, Evans, Myer, Gwin, Heidt Jr, & Hewett, 2006; Martinez, Dura, Gamez, Rosa, Zamora, & Alcantera, 2004). Further still, these studies have only compared the surfaces at one time of the year. Season variations in climate and use may influence the properties of natural turf and third generation artificial surface and particularly affect how the performer responds to the surface. Likewise, there is little research literature available that details the biomechanical response of the athlete to changes in construction of third generation artificial turf surfaces.

Dixon et al. (2008) also performed research into biomechanical differences associated with different soccer boots on natural surfaces constructed in a soil bin. The authors observed significant differences between soccer boot footwear conditions, although the differences depended on the surface cushioning. Thus, the response to soccer boots may change depending on the surface played. As such, consideration of surfaces is important when evaluating soccer boots. Additional changes to the boot have been shown to prevent injury in running and soccer playing populations. These have included changes to the footwear cushioning and an increase in the heel height of the athlete, yet the mechanisms behind such changes in injury prevention are unclear. Likewise, it is not known whether these changes will be in contrast to the design consideration of the soccer boot and thus predispose the soccer player to an alternative injury. Therefore, the limitations of previous surface and footwear studies should be addressed in order to further understand the aetiology of injury in soccer and appreciate the effect of risk factor changes on the prevention of injury. As such, the specific aim of this thesis to address the research question *"what effect do the playing surface and footwear variables have on the biomechanical response of soccer players"*.

To achieve this overall objective, each experimental chapter has individual aims. The first experimental chapter aims to assess the extent to which biomechanical characteristics change when soccer players run and turn on natural turf and third generation surfaces at different times of the year which differ in environmental conditions. The second question in chapter 3 (study 1) is concerned with the extent to which the biomechanical characteristics of soccer players change in different soccer boots and how do the boots interact with the surfaces to influence the biomechanical characteristics. The final research question asks to what extent do biomechanical characteristics change on natural and third generation artificial turf surfaces between two contrasting times of the year.

In chapter 3, biomechanical observations are made on soccer players when running and turning on 'real' outdoor natural and a third generation artificial surface. Because of the potential change in natural turf over time, the comparison was conducted at two times of the year. These times coincided to when the climate was expected to be warm and dry and cold and wet, in an attempt to alter the mechanical properties of the surface. Likewise, because the choice of footwear is implicated in the occurrence of soccer injuries, different boots were tested on each surface along with any interaction effects observed between the playing surface and footwear variables. Further still, the biomechanical measurements were used to assess the extent of any movement characteristic change that occurs on the natural and artificial surface between two test occasions. This research aims to provide important and novel information into the biomechanical characteristics of participants that use third generation artificial turf and natural turf and the effect that the time of year may have on the results. This contributes to a greater understanding of the biomechanics of soccer players on different playing surfaces and footwear. It may also provide information into the relationship between hard playing surfaces and the high incidence of injury compared to periods of the year where the surfaces are more cushioned and how these injuries may be reduced.

In chapter 4 (study 2), it is acknowledged that despite the findings presented in chapter 3, the response of the soccer player may differ if the construction of the third generation artificial changes. In particular, the shock pad layer positioned beneath the third generation artificial turf carpet can have different densities and construction. These differences are purported to influence the cushioning provided to the participant, yet there is little biomechanical evidence to support this speculation. Thus, in chapter 4, measurements are made with the aim of assessing the biomechanical cushioning characteristics of soccer players when running and turning on two different shock pad densities.

The placement of a heel lift in footwear has been shown to lower the risk of sustaining overuse injuries (Faunø, Kålund, Andreason, & Jørgensen, 1993; MacLellan, 1984; MacLellan & Vyvyan 1981). However, the mechanism behind such injury reduction is unclear. This ambiguity can be related to the effect that the movements performed, the type of footwear worn, and the biomechanical measurements used have on the accuracy of the results. In chapter 4, these limitations are addressed in order to improve the current understanding of the mechanisms behind injury reduction. Furthermore, cushioning insoles have been shown to reduce lower extremity pain (Tooms, Griffin, Green, & Cagle, 1987) and biomechanical measurements associated with injury (Dixon, Waterworth, Smith, & House, 2003; House, Waterworth, Allsop, & Dixon, 2002; Rööser, Ekbladh, & Lidgren 1988; Windle, Gregory, & Dixon, 1999). Sorbothane® claim to have made a cushioning insole for specific use in soccer which can lower '*much of the vibration created every time your foot hits the ground*' which '*takes the pressure off your feet'* and '*can significantly reduce much of the leg and back pain associated with hard exercise*' (Sorbothane®, Appendix A). However, their claims are made without any published scientific support. The second experimental chapter in chapter 4 will therefore endeavour to understand the biomechanical mechanisms behind the successful application of heel inserts in lowering the risk of overuse injury and the effect soccer specific insoles can have on the biomechanical behaviour of soccer players.

The final question addressed in this chapter (Study 2), focuses on the extent to which the inclusion of cushioning insoles and heel inserts have on the risk of over lower extremity injuries. The inclusion of a heel insert and cushioning insole was assessed for the risk of sustaining an acute ankle injury. As such, external biomechanical measures associated with ankle ligament were taken. The evidence provided in this chapter, contributes to the understanding of how the construction of artificial surfaces can influence the performer. Such information may potentially be used to reduce the risk of future injury. The evidence from the footwear changes also aims to provide biomechanical evidence into how the methods have been successful in reducing injury, and which can be used to assess changes in the future design of soccer boots.

In chapter five (Study 3), the evaluation of the heel lift intervention is continued by considering its ability to reduce Achilles tendon force when running and turning. To achieve this, three dimensional plantar flexion joint moments and an estimation of Achilles tendon force were collected, along with the average loading rate of each measurement. The calculation of the Achilles tendon force was also made using three dimensional plantar flexion moments and a dynamic three dimensional moment arm distance from the ankle joint centre to the line of the Achilles tendon. This is important for the accuracy of the calculation due to the three dimensional nature of human movements. This also improves on the limitations of previous methods which used sagittal plane moments and moment arms. These measurements were performed during running and turning movements whilst wearing a soccer boot. This provides a greater understanding regarding the effect of the movement type on the loading of the Achilles tendon. Dorsi-flexion and eversion joint moments and average loading rates of these measurements are also used to assess whether soccer players are at a greater risk of acute ankle injury with the heel lift during a turning movement. This aims to provide a more detailed insight into how Achilles tendon injury is reduced with the heel insert. This may also provide appropriate measurements in which future footwear design changes can be assessed. However, this data only provides the specific measurements that should be used alongside prospective injury studies to link the reduction in these values with lower injury rates. Heel inserts may be used in soccer, particularly at times when the incidence of injury is high. However, any change in the data associated with the risk of ankle ligament damage may raise questions regarding the suitability of the device for use in soccer.

In addressing soccer injury risk factors in the current thesis, it is hoped that appropriate recommendations can be made into methods of injury avoidance in the future. Therefore, the key findings and subsequent recommendations are summarised and discussed in chapter 6. This chapter details the implications of the findings to the larger research community as well as practical recommendations regarding how soccer injuries, particularly during preseason and early in-season may be avoided. It also indicates which surfaces to use at a given time. The playing surface may also influence the choice of footwear as well as the effect that additional cushioning could have to protect the lower extremity. As such, there could be a reduction in the preseason injury rates as well as the in-season injury rates that occur as a result of re-injury. This may then contribute to a reduction in the number of injuries compared to other sports. Importantly, this may have a positive financial, emotional, physical and psychological consequence for the soccer player.

The findings of this thesis will provide information into potential methods by which injury risk may be prevented and also provide an insight into the mechanism behind injury risk and reduction already present in soccer. In providing this information it may be possible to inform soccer clubs, governing bodies or even individual players of ways in which injury may be avoided. This may help lower the overall incidence of injury and reduce the financial, emotional, psychological and physical effect of injury on the soccer player. It may also be useful to sports that also use similar footwear and surfaces, such as rugby, Australian rules and American football.

2. Review of Literature

2.1 Introduction

The initial purpose of this chapter is to provide an introduction to the origins and popularity of soccer (section 2.2) followed by a description of the injury problems associated with participating in this sport and the implications these injuries have for the players (section 2.3). The effect of seasonal climate variation is also discussed in this section, where it is acknowledged that the time of year influences the number and severity of injuries that occur in soccer.

In trying to lower the total incidence of injury, it has been suggested that the aetiological risk factors that predispose the player to an increased risk of injury and the causal mechanisms, must first be identified (Bahr & Holme, 2003). In section 2.3.2, a description of the risk factors associated with soccer injury is given, followed by section 2.3.3, which details the different mechanisms through which an injury can occur. These can be divided into those that are sustained through overuse (chronic) and those that occur acutely. Further still, the identification of the mechanism behind a specific injury may aid the identification of the correct risk factors for a specific injury. Therefore, it is imperative that the correct mechanism behind the injury is identified prior to identifying the key risk factors and possible injury avoidance interventions. Because of this, two overuse injuries and one acute injury common in soccer, along with their associated risk factors, are detailed in section 2.3.4.

To assess the effect of different risk factors, biomechanical measurements are often used. This can include external kinetic and/or kinematic measurements, and also an estimation of internal joint loading. Measurements of kinetic variables are associated with external forces, and are linked with impact and propulsion related lower extremity injury. These forces are discussed in detail in section 2.3.5.1 along with the different methods by which these variables can be measured. Kinematic data is another external measurement that can be collected and used to analyse change in the body's geometry when participants experience the different risk factors. In section 2.3.5.2 the importance of kinematic adaptation with reference to attenuating forces is identified and the technique of collecting kinematic data is described. The key joints of the ankle-foot complex are also outlined in this section, as is how the foot-ankle complex can be measured alongside the movements of the knee to evaluate the risk of lower limb injury. Finally, in section 2.3.5.3 a thorough explanation is given regarding the calculation of joint moments, the estimation of the forces that occur internally, and the potential role that joint moment measurements have in understanding injury causation.

Care must be taken when investigating the biomechanical adaptation to risk factors as the response can differ depending on the movement performed. Soccer is a multi-directional sport and the task direction can have implications on the type and location of the injuries that occur. The choice of activity can influence the loading of specific structures differently. Consequently, a risk factor may not influence injury during one task but may do during another. Therefore, when testing a risk factor, the measurement of biomechanical variables may not be as revealing under some situations as during others. As such, the effect of different movements on the assessment of various risk factors is reviewed in section 2.3.6.

Using the knowledge of biomechanical measurements and the influence that the correct movements can have on results, a critical analysis is given in section 2.4. This uses previously published literature which have highlighted any biomechanical changes that occur with different natural playing surfaces (section 2.4.1) and soccer specific footwear (section 2.4.2) which are linked to the injuries presented in section 2.3.4.

In understanding the risk factors associated with injury in soccer, it may be possible to adapt the risk factors and lower the potential risk. In section 2.5, discussion is given regarding the techniques already used in other sports (or situations) to lower the injury risk, alongside biomechanical changes that occur. One currently prescribed method is the insertion of additional material inside poorly cushioned footwear. This has been shown to reduce the risk of injury and alter certain biomechanical variables. The available evidence and the application of such material to soccer boots are described in section 2.5.1. However, in changing the interface between the boot and the foot when using a shoe insole/insert there may be a potential for increased risk of other injuries. This important aspect is also discussed in this section.

The potential for risk factor adaptation may also be observed if the natural playing surface was changed to an artificially constructed one. In section 2.5.2.1 there is discussion regarding the introduction of the traditional artificial surfaces and their use in modern soccer. There is also detail of previous comparisons of natural and traditional artificial turf surfaces regarding the associated change of both the incidence of injuries and the biomechanical response of participants'. Third generation or "in-fill" artificial surfaces are then evaluated in section 2.5.2.2, where the rate of injuries to players using this soccer specific playing surface are compared to natural turf surfaces. Additionally, mechanical and biomechanical studies which compare this artificial turf to natural turf, and the effect that changes made to the composition of the third generation artificial turf have on the participant, will also be addressed.

The response of the participant to the surface and footwear conditions has been associated with kinematic changes, which enable impact forces to be maintained across differently cushioned surfaces. Literature detailing these kinematic changes is described in section 2.6. Finally, the review of literature is summarised and concluded in section 2.7, where the aims and hypotheses for the experimental chapters included in this thesis are stated.

2.2. Soccer: A brief history

Although many forms of football have appeared in historical text from as early as 3rd century BC, and as far away as China, the origins of modern football can be traced back to the public schools of 19th century England (Blain, 2009). Initially, there were no generic rules. Instead, the regulations differed according to the school in which the game was played. When football was popularised by the English masses, leagues were set up and rules were standardised. However, when one hugely conflicting rule was discussed, there was a failure to establish a compromise. On one side of the argument teams welcomed the use of the hands and feet by all players, whereas on the other side, teams wished only one specified player the privilege. As a consequence, football was split into two codes, Rugby Football and Association Football, and these were played separately from one another.

In 1863, the Football Association (FA) was set up to govern the rules of Association Football in England (The FA, 2009), followed by the Fédération International de Football Association (FIFA), (FIFA, 2009) in 1904 to oversee the game across the world. Football as it is officially called by FIFA and is commonly referred to in Europe, is otherwise known as "soccer" across the rest of the world because of the soc. in association (Blain, 2009). This allows the game to be distinguished from rugby and the other forms of football that has been developed since.

Today, soccer has been described as one of the most popular sports in the world with approximately 200,000 professional and 240 million amateur players registered with FIFA (Judge & Dvorak, 2004). These participate in approximately 204 countries (Andersen, Larsen, Tenga, Engebretsen & Bahr, 2003). Soccer is also played across many different age groups (Inkaar, 1994) and is the most watched spectator sport in the world (Andersen, et al., 2003). There are also many variations of soccer, differing in the number of participants, construction and size of playing surfaces, footwear designs, game length and size of playing equipment (balls and goals). Despite these differences, the aim of soccer is generally the same; that is all players except the goal keeper must try to propel a round football without using their hands or arms, into the opponent's goal. The most traditional form of soccer involves two teams of 11 players, participating in a game that lasts 90 minutes, consisting of two 45 minute halves, with a 15 minute rest interval (Wong & Hong, 2005).

Soccer is a complex contact sport (Inkaar, 1994) that demands physical, physiological, technical, and tactical skills. Soccer is a dynamic sport that requires multi-directional manoeuvres and is characterised by sprinting, stopping, cutting and pivoting situations (Inkaar, 1994). Soccer also requires the athlete to perform intermittent multi-intensity exercise, ranging from low-levels during walking and jogging during recovery time, to high intensity sprinting (Abrantes, Maçãs and Sampaio, 2004; Wong & Hong, 2005). Elite soccer players are also aerobically fit, running an average of 11km in 90 minutes (Bangsbo, 1994).

2.3 Soccer injuries

The risk of injury is present in all sporting activity. However, prospective and retrospective research studies, have described the risk of injury in soccer as 'considerable' (Andersen et al., 2003), and 'high' compared to other sports (Hawkins, Hulse, Wilkinson, Hodson, Gibson, 2001; Junge et al., 2004; Weightman & Browne, 1975; Yde & Nielsen, 1990). Certain risk factors can make soccer more dangerous than some high-risk industrial occupations such as agriculture and construction (Drawer & Fuller, 2002; Hawkins et al., 2001).

Suffering an injury can have physical, psychological and financial consequences for the participant (Fuller & Drawer, 2004; Pritchett, 1981; Woods et al., 2001; Rahnama, Reilly & Lees, 2002). For example, a loss of playing time can cause a reduction in revenue from supporters wanting to see the best players (Wong & Hong, 2005). Teams may also fail to do as well in competitions, which can directly influence the prize monies won and further discourage supporters from watching their team (Woods et al., 2001). Furthermore, if the participant continues to train and/or compete with an injury their performance could be impaired (Fuller & Drawer, 2004), which could lower the chance of being successful. Also, if these injuries are repeatedly sustained, they can have long-term effects on the participant's mental and physical well-being (Fuller & Drawer, 2004). In the most extreme case, multiple minor injuries can lead to a more serious injury and ultimately result in the player having to retire early (Drawer & Fuller 2001; Fuller & Hawkins, 1997). In professional soccer this could cause a player to suffer a loss of income, which may be unexpected. As such, players may be financially un-prepared for this loss, which could cause an added burden to go along side the trauma of injury. The financial cost of injury may even be observed by semi-professional or amateur soccer players as they may not be able to work in their regular occupation (Wong & Hong, 2005). Treatment of injury is also a huge financial burden on the health service (Pritchett, 1981). Cullen & Batt, (2005) reported that musculoskeletal injury treated by the National Health Service costs approximately £590 million per year. Sports and exercise contributes significantly to this high cost of treatment. Serious injury can also affect their long term health of both professional and amateur soccer players as they may be unable to participate in an active lifestyle if the injury becomes untreatable. Therefore, the pain and discomfort, along with the frustrations that come from the rehabilitation of injury, and the high financial cost, provide clear reasons for research into the understanding and prevention of injury risk in soccer.

Van Mechelen, Hlobil & Kemper (1992) presented a four-step paradigm by which the prevention of injury could be addressed.

Step 1: To establish the extent of the sports injury problem
Step 2: To establish the specific aetiology (risk factors) and mechanisms of injuries
Step 3: To introduce preventative measures
Step 4: To assess the effectiveness of the preventative interventions by repeating step 1

Steps 1 and 2 of this framework are used in this literature review to discuss the existing research that relates to the occurrence and severity of certain soccer injuries. It is also used to highlight the biomechanical characteristics of soccer players whilst experiencing the various risk factors. In addition, the framework is used to identify research that is needed to further knowledge behind the characteristics of soccer players when these injuries potentially occur, and the interventions presently prescribed to reduce similar injuries in other sports.

2.3.1. Establishing the extent of the sports injury problem in soccer.

The first step towards the prevention of injury is to establish the extent of the injury problem. Although soccer injury has been described as considerable (Andersen et al., 2003), particularly compared to other sports (Hawkins et al., 2001; Junge et al., 2004; Weightman & Browne, 1975; Yde & Nielsen, 1990), the term soccer injury encompasses all injuries that occur during participation which vary in terms of their anatomical location and severity. Therefore, the overall injury value does not provide enough detail into which injuries are the most common, and as such is not particularly useful when trying to lower the occurrence of injury. Instead, it is more valuable to identify the specific, most problematic injuries which are the largest contributor towards to the high rate of injury found in soccer. As such, it may be possible to apply suitable interventions to lower the risk of these specific ailments.

Due to the bipedal nature of soccer, many injuries occur to the lower extremity. Across this region, the ankle and knee joints are most commonly injured (Chomiak et al., 2000; Ekstrand, & Gillquist, 1983; Nielsen & Yde, 1989; Shea, Pfeiffer, Wang, Curtin, &

Apel, 2004; Waldén et al., 2005a; Woods et al., 2002), with muscle and tendon strains, and ligament sprains, being consistently reported (Arnason, Gudmundsson, Dahl, & Jóhannsson, 1996; Ekstrand, & Gillquist, 1983; Wong & Hong, 2005; Woods et al., 2002). However, although these sites are most common, the risk of injury to these regions can vary throughout the year.

In Europe, the soccer season traditionally starts in mid-August and ends in May (Woods et al., 2002). This is then commonly followed by the “closed season”, which can last between 2–5 weeks. Players subsequently return for 4–6 weeks of preseason training during July and August in preparation for the beginning of the competitive season (Waldén et al., 2005a; Woods et al., 2002). These are summer months and during this time the weather has been described as being mainly dry and warm with low humidity (Meyers & Barnhill 2004; Woods et al., 2002).

In sports such as soccer that start in summer and early autumn, the risk of injury can increase during this time. Orchard (2002), Woods et al. (2002) and Woods et al. (2003) observed the presence of an early season bias towards certain injuries, which is not typically reported in summer football competition or indoor sports such as basketball. These injuries are generally less severe than those experienced at other times, with a significantly greater percentage of slight and minor injuries being observed (Woods et al., 2002). However, it has been shown that the experience of an initial injury will put the participant at a considerably greater risk of re-injury at a later date (Arnason et al., 1996; Chomiak et al., 2000; Ekstrand, & Gillquist, 1983; Woods et al., 2002). Indeed, one third of soccer players who experience a minor injury in the first instance, sustaining a more serious injury at the same location (Arnason et al., 1996; Chomiak et al., 2000; Ekstrand, & Gillquist, 1983). Likewise, in response to the first injury, participants may change their movement patterns. This can put other previously less used structures at an increased load and unexpected stress, accentuating the risk of a new injury to result. Therefore, much of the high injury rate observed in soccer may be reduced by establishing the causes of the most common injuries during the preseason period.

2.3.2. Aetiology (risk factors) of soccer injuries.

Although a considerable number of injury-related studies have described the incidence, type, location, severity and pattern of soccer injuries, much less is known about the risk factors and mechanisms that cause injury. The correct identification of these factors is fundamental for any successful injury prevention strategy (Anderson et al. 2003; Van Mechelen et al., 1992). Risk factors are the aspects of the sport that increase the susceptibility to injuries. Traditionally, risk factors are divided into those that are internal to the athlete and those that are external to the athlete (Gissane White, Kerr & Jennings, 2001). However, Bahr and Holme (2003) suggested that it is more appropriate to describe risk factors as those that are modifiable and those that are non-modifiable. The contribution of each can change depending on the injury location, and in most cases injury causation is multi-factorial. As such, injury often results from a combination of internal risk factors (e.g. player characteristics), external risk factors (e.g. environmental and equipment characteristics), and injury mechanisms (Andersen et al., 2002).

Non-modifiable factors associated with an increased risk of lower limb ailments in soccer players include age (Chomiak et al., 2000; Shea et al., 2004; Wong & Hong, 2004; Woods et al., 2002), gender (Arendt & Dick, 1995; Lindenfeld, Schmitt, Hendy, Mangine, & Noyles, 1994; Shea et al., 2004; Wong & Hong, 2004), and anthropometric characteristics (i.e. muscle tightness, limb length etc) (Messier & Pittala, 1988). Those factors described as external and modifiable can include playing surfaces, footwear, training intensity, and the quality and length of rest time (Chomiak et al., 2000). In most cases only one or two factors are examined, rather than testing the effect of multiple risk factors, which may interact to bring about a given injury. Likewise, despite considerable interest in internal and non-modifiable factors, these are often innate to the athlete, and cannot be easily changed or standardised for groups of participants. Instead, the focus of many injury prevention studies has been on those described as external and modifiable through physical training or behavioural approaches (Bahr & Holme, 2003).

2.3.3 Injury mechanisms.

It has been suggested that the mere presence of a risk factor is not sufficient to cause injury, but only renders an athlete susceptible (Bahr & Holme, 2003; Meuwisse, 1994). To establish a complete understanding, the causal mechanisms are needed (Bahr & Holme, 2003; Bahr & Krosshaug, 2005; Meeuwisse, 1994; Van Mechelen et al., 1992). Classification of an injury mechanism in the first instance can be divided into those that are caused by contact between the athlete and another player or equipment, and those that occur without contact and are caused by the athlete alone. In soccer, non-contact injuries are more common than contact injuries at most anatomical locations (Wong &

Hong, 2005), and occur mainly during running and turning movements (Junge & Dvorak, 2004). Hawkins and Fuller (1999) argued that understanding the mechanisms behind non-contact injury is an important area for consideration if the incidence of injury in soccer is going to be reduced.

Non-contact injuries can be grouped into those that are overuse or chronic and those that are acute or traumatic. Overuse injuries can be characterised by pain across the musculoskeletal system and result from repetitive trauma with insidious or subtle onset, which prevents adequate tissue repair (Bartlett, 1998; Waldén et al., 2005a). These injuries occur without any known trauma or disease that might have explained the previous symptoms (Waldén, Hägglund, & Ekstrand, 2005b). On the other hand, traumatic injuries are defined as being acute (Waldén et al., 2005b), and are often a result of one sudden and traumatic experience (Bartlett, 1998), such as a sudden twist or fall whilst running and/or turning. In this case, the force that is produced is in-excess of the tolerance level of the structure and so immediate injury may follow. During pre-season, as well as early in-season, overuse injuries are most common (Woods et al., 2002). The comparative number of overuse injuries are also greater during this time than at any other (Ekstrand & Gillquist, 1983; Hawkins et al., 2001; Woods et al., 2002; Waldén et al., 2005a) and are being described as disproportionately ‘high’ (Hawkins et al., 2001). However, when looking at all the injuries that occur in soccer, independent of the time of year, acute ankle sprains are the most common (Waldén et al., 2005a).

Care must be taken when trying to classify an injury as overuse or acute as it is possible to suffer an injury to some regions or structures, such as the ankle and knee, by either overuse or acute mechanisms (Ekstrand & Gillquist, 1983). One example of this is the structure of the Achilles tendon (McLauchlan & Handoll, 2001; Schiffer, 2006). The aetiological factors for most acute and overuse injuries are similar and result in variations in the speed of the symptoms (McLauchlan & Handoll, 2001). These symptoms include pain and discomfort or swelling and inflammation, and determine whether a warm up will help prevent the injury (McLauchlan & Handoll, 2001) and which risk factors are influential (Saglimbeni & Fulmer, 2009). Thus, if the first injury was prevented, the susceptibility to the second injury may be avoided. Failure to establish the correct risk factors can mean that the treatment of an injury, may at times be unproductive or even potentially harmful to the participant. Therefore, for the development of a successful injury avoidance strategy, specific injuries must be identified along with the correct injury mechanisms in order to identify the correct injury risk factors.

2.3.4 Soccer specific injuries and the associated risk factors.

2.3.4.1. Achilles tendon injury

Various statistics have demonstrated that of all overuse injuries, the Achilles tendon is one of the most frequently injured structures in the body (Nigg, 1986). Achilles tendon injury is especially problematic in ball games such as soccer (Kvist, 1994; Paavola, 2001; Paavola, Kannus, Järvinen, Khan, Józsa, Järvinen, 2002; Mazzone & McCue, 2002; Waldén et al., 2005a). Further still, Woods et al. (2002) observed that nearly one third (32%) of overuse related Achilles tendon injuries were sustained during the preseason period (94% were either tendonitis or paratendonitis). Achilles tendonitis is caused by damage and inflammation to the tendon (Paavola et al., 2002; Mazzone & McCue, 2002) and is particularly susceptible to injury because of it structural and functional demands (McLauchlan & Handoll, 2001).

The triceps surae muscle group (gastrocnemius and soleus) is the largest and strongest muscle complex in the calf (Mazzone & McCue, 2002) and spans two joints via the Achilles tendon, to connect to the heel (calcaneous) (Schiffer, 2006) (Figure 2.1.). Most Achilles tendon problems occur about two inches above the heel bone, at an area of the tendon that has relatively poor blood supply (Schiffer 2006). This structure renders the tendon complex extremely vulnerable to injury (Mazzone & McCue, 2002; Saglimbeni & Fulmer, 2007; Schiffer 2006) especially when subjected to strong forces (Mazzone & McCue, 2002). The poor blood supply also accounts for the relatively long time taken for the Achilles tendon injury to heal (Schiffer, 2006).

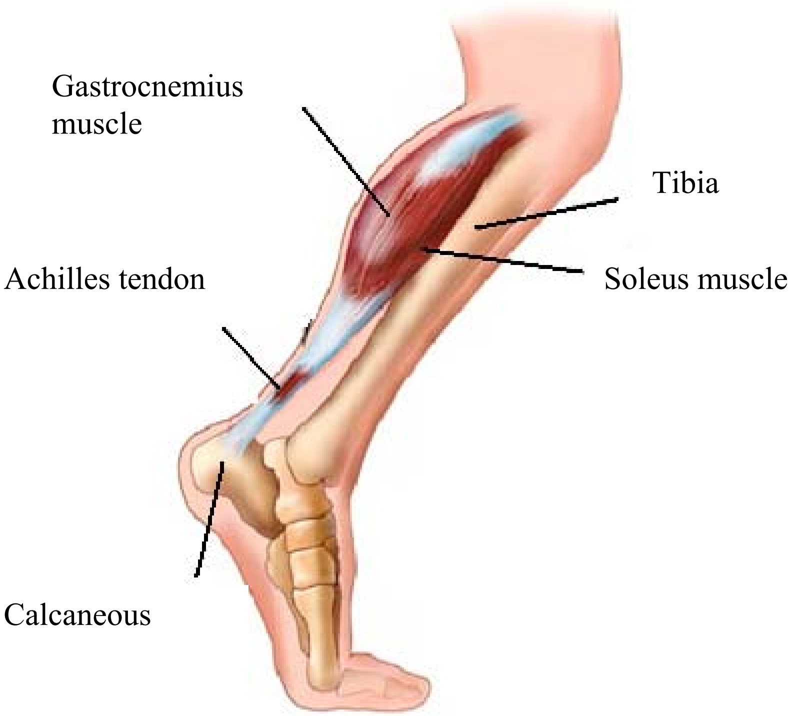


Figure 2.1. Location of the Achilles tendon, in relation to the Gastrocnemius and Soleus muscle groups and the Tibia and Calcaneous (Adapted from Takano, 2009 with permission of the author)

The Achilles tendon transmits the force of the triceps surae muscles during concentric contraction to provide ankle plantar flexion (Whiting & Zernicke, 1998). During running and walking the gastroncemius produces force during mid-stance of gait, which is applied through the Achilles tendon (Komi, 1990) to help prevent the downwards transition of the centre of mass (COM). During tasks such as running, the Achilles tendon experiences a lengthening and shortening of the Achilles tendon complex in response to differences in ankle angle during the gait cycle. This causes stress, strain and extreme shear forces to be generated. In response to these forces, micro trauma develops that is not always felt or seen by the participant, providing sufficient rest time is given for the tendon to regenerate (Mazzone & McCue, 2002). This serves as a necessary mechanism by which the Achilles tendon can thicken and become stronger. Magnusson and Kjaer (2003) showed that the cross sectional area of the Achilles tendon was greater in runners than in non-runners. This was suggested to indicate a region-specific hypertrophy in response to habitual loading during running and was put forward as a mechanism by which the Achilles tendon stresses were reduced, thereby lowering the risk of injury.

The reduction of load tolerance and subsequent susceptibility to overuse Achilles tendonitis (tendonopathy) has also been shown to occur when micro-trauma develops in response to increased sub-maximal eccentric loading of fatigued muscle (Clement, Taunton & Smart, 1984). This eccentric load occurs as the ankle dorsi-flexes during the mid-stance of the gait cycle. With such an increase in sub-maximum load, injury is not immediate, but results when sufficient rest is not given for the structure to recover from this micro-trauma. The effect of Achilles tendon damage can then become observable when the loading magnitude may no longer be endured by the participant (Renstrom, 1994) which depending on the severity of damage causes a reduction in performance.

The mechanical theory of “tendon overuse” states that when the tendon has been strained repeatedly to approximately 4-8% of its maximum tension, it is unable to endure further tension and damage begins (Paavola, 2001). This is because the tendonous tissue becomes fatigued as the ability of the tendon cells to repair the fibre damage is over whelmed by repetitive micro-trauma. The structure of the tendon is then disrupted both at the micro- and macro-scopic levels (Paavola, 2001) by this repetitive strain. Collagen fibres can begin to slide past one another, causing breakage to their cross-linked structure and denature, which results in inflammation, edema and pain (Jozsa & Kannus, 1997 cited in Paavola, 2001). This cumulative micro-trauma and inflammation is thought to weaken the collagen cross-links, the non-collagenous matrix and vascular elements of the tendon, and finally lead to Achilles tendonopthy. Leadbetter (1992) termed this the tendonosis cycle. With repeated sub-maximal tension of 4-8%, chronic Achilles tendoinosis can occur, which depending on the length of time and number of repetitions, can be characterised by degenerative changes with symptoms that last a considerable length of time, even when conventional non-surgical treatment is used (Schiffer, 2006). Also, after chronic inflammation, degeneration can occur where the tendon becomes weakened. Therefore, the tendon cannot respond to forces that were previously within the tolerance range. In these cases, a more severe Achilles tendon injury has been found to occur in conjunction with the previous Achilles tendon injury. This can be observable as a Achilles tendon rupture (Kannus & Jozsa, 1991 cited in Paavola, 2001; Saglimbeni & Fulmer, 2007), although a direct causal relationship has yet to be established (Kannus & Jozsa, 1991 cited in Paavola, 2001). This process may therefore describe the way in which overuse Achilles tendon injury occurs in soccer. However, it does not explain why the injury rates are greater during preseason.

The development of overuse Achilles tendonopathy is a multi-factorial process influenced by both external and intrinsic risk factors which can include biomechanical and non-biomechanical factors (Williams, 1986; Van Ginckel, Thijs, Hesar, Mathieu, De Clercq, Roosen & Witvrouw, 2009). An increased sub-maximal running speed or jumping movement can accentuate the loading of the Achilles tendon and increase the risk of injury (Mazzone & McCue, 2002). Biomechanical factors include underlying anatomical features (Van Ginckel et al. 2009; Mazzone & McCue, 2002; Paavola, 2001), such as malalignment of the foot (hyperpronation and hypopronation), have been linked to the risk of Achilles tendon injury (Paavola, 2001). This is especially the case for hyperpronation (excessive eversion, dorsi-flexion and internal rotation) which is considered to be the most pertinent hypothesis concerning the relationship between running patterns and injury (Van Ginckel, et al. 2009; Clement et al., 1984).

Achilles tendon injury risk factors also include hard playing surfaces (Wood et al. 2002; Saglimbeni & Fulmer, 2007), inadequate or inappropriate training shoes (Wood et al. 2002; McLauchlan & Handoll, 2009; Saglimbeni & Fulmer, 2007), poor training technique (McLauchlan & Handoll, 2009), an imbalance between physical load and load tolerance of the athlete (Woods et al., 2002) and a sudden rise in exercise intensity (Woods et al., 2002; Smart et al., 1980; Saglimbeni & Fulmer, 2007). Collectively these are factors which are associated with the disproportionate number of Achilles tendon injuries that occur during pre-season in soccer (Woods et al., 2002). This is because these factors exert the greatest influence over the loads experienced by soccer players during the pre-season period. The other factors influencing Achilles tendon injury are often innate characteristics of the participant (i.e tight gastrocnemuis or Achilles tendon) and which will always influence the risk of injury, irrespective of the time of year. However, if soccer players possess these risk factors, their effects are likely to be accentuated by differences in the time of year.

4.3.4.2. Lateral ankle ligament injury.

Ankle injuries are most common in sports that involve multi-directional running, require good balance, and entail quick stop-start movements (Pollard, Sim & McHardy, 2002). These requirements are common in soccer, and as such, so is the risk of ankle injury (Ekstrand & Gillquist, 1983; Engebretsen, Myklebust, Holme, Engebretsen, & Bahr, 2008; Pollard et al., 2002). In fact, of all the injuries that occur in soccer, acute ankle damage is the greatest. However, a greater proportion of these ankle injuries are caused by contact. Thus, the risk factors associated will be different.

As was observed with overuse injuries, the time of year has also been shown to influence the risk of ankle ligament damage. An injury bias towards warm and dry conditions has been reported in sports such as Australian rules and American football, and rugby union, occurring during the pre-season and early in-season (Orchard, 2002; Orchard, 2001; Orchard & Powell, 2003; Lee & Garraway, 2000). In soccer, a similar early season bias has been observed. Woods et al. (2003) found that during the first three months of the season including July and August which coincides with pre-season, 44% of the yearly total of ankle injuries were sustained, which was significantly greater than the rest of the year. Player to player contact was responsible for 59% of the ankle injuries, and 39% were non-contact injuries. Even though the proportion of non-contact injuries is smaller, it has been recommended that to lower the total injury risk in soccer, non-contact injuries mechanisms should be evaluated (Hawkins & Fuller, 1999). Likewise, this magnitude is still large and thus would have a considerable effect towards the reduction of injury in soccer. Focus should therefore be given towards these injuries. During the early season the environmental conditions are warm and dry (Woods et al., 2002). These environmental conditions influence the playing surface hardness and contribute towards the shoe-surface traction, which are associated with non-contact ankle injuries in sports such as soccer (Orchard, 2002).

It has been well documented that previous injuries are a major risk factor for re-injury amongst professional soccer players (Dvorak, Junge, Chomiak, Graf-Baumann, Peterson, Rösch & Hodgson, 2000; Nielsen & Yde, 1989; Arnason, Sigurdsson Gudmundsson Holme, Engebretsen & Bahr, 2004). This is particularly the case for the ankle, where 56% to 75% are re-occurring injuries (Woods et al., 2003). This predisposes a player to further, more severe ankle sprains later in the season, that often occur as a result of a non-contact mechanism (Woods et al., 2003). Ekstrand and Gillquist (1983) reported that occurrence of these major injuries were related to an impairment of timing and neuromuscular coordination resulting from the initial injury. By understanding and avoiding the initial conditions in which non-contact injury can occur, initial injury may be avoided. This may lower the subsequent susceptibility to a more severe re-injury, possibly caused through a non-contact mechanism, which may then contribute to a lowering of the total injury risk observed in soccer.

During quick stop-start, multi-directional tasks such as turning, cutting and side-stepping, most movement and loading occurs on the lateral side of the foot. Consequently, these are often classified as lateral movements, and the magnitude of this loading can influence the alignment of the lower extremity, determining the stability of the joint and the risk of athletic injury. Stability of the ankle joint is provided by the interaction of the ligamentous structures (Pollard et al., 2002), which act to bind the many bones of the foot together to reinforce the ankle and prevent excessive movement (Norkin & Lavangie, 1992). There are also many ankle muscles that are activated prior to and during the early stance phase which help stabilize the ankle and foot during rapid motion (Nakazawa, Kawashima, Akai & Yano, 2004). However, when a rapid deceleration process is undergone, ankle stabilisation may not always be possible. This can cause the maximum ankle movement range to increase and the joint system of the foot and ankle to experience higher loads. This may result in pain and injury to the joints and the associated structures (Tiegermann, 1983). This rapid deceleration may account for the high frequency of ankle ligaments sprains found in soccer.

The ankle ligaments that are at risk of injury exist as two separate groups. Firstly, there are the anterior and posterior tibio-fibular ligaments that aid in maintaining the grasp of the tibio-fibular mortice on the body of the talus (Pollard et al., 2002). Secondly, there are those ligaments that maintain contact with the ankle joint surface and control medial and lateral glide. These include the medial and lateral collateral ligaments (Pollard et al., 2002) (Figure 2.2). These ligaments are primarily responsible for restricting inversion/eversion and dorsi-flexion/plantar flexion movements of the ankle (Whiting & Zernicke, 1998). The range of inversion during lateral movements is an important factor when assessing ankle injury. Baumhauer, Alosa, Renstrom, Trevino & Beynnon, (1995) found that 85% of all ankle injuries are caused by excessive inversion trauma, which causes sudden lateral forces that are sufficient to compromise the joint integrity. Seventy-five percent of all ankle injuries were ligamentous (Fallat, Grimm, & Saracco, 1998) and 75% of ankle injuries are acute ankle sprains (Barker, Beynnon & Renstrom 1997). Therefore, a link can be established between inversion trauma and ankle ligament sprains.

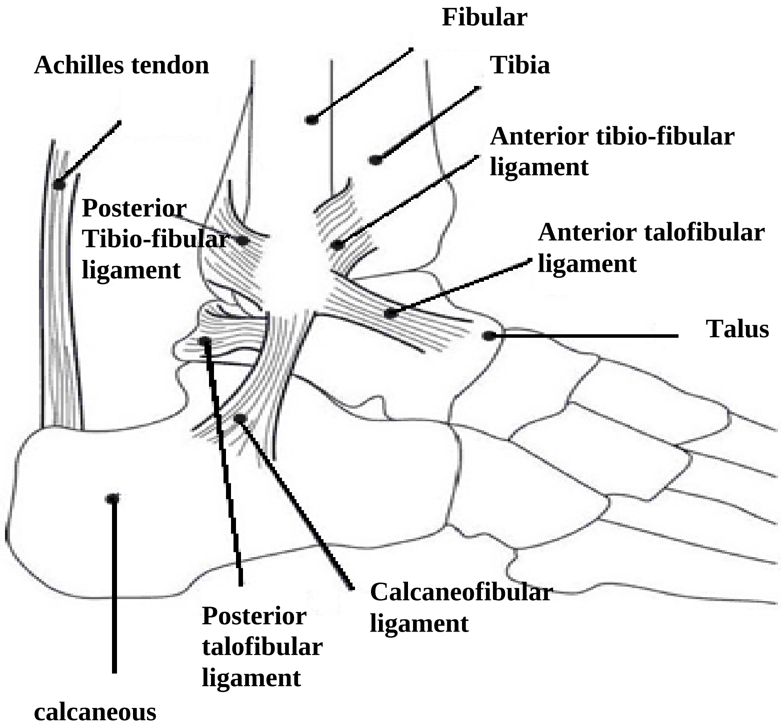


Figure 2.2. Diagram of the ankle joint structure and ligaments (Adapted from U.S. Department of Health and Human, 2009).

Woods et al. (2003) reported that 77% of ankle sprains in soccer occur at the lateral ligament complex. Within the group of lateral collateral ligaments, the anterior talofibular ligament (ATFL) works along side the calcaneo-fibular ligament (CFL) to resist ankle inversion, but where the CFL resists inversion in a dorsi-flexed position, the ATFL restricts the range of inversion whilst the ankle is in the plantar flexed position (Pollard et al., 2002). In soccer, injury to the ATFL is thought to occur in 66% - 73% of the cases of ankle sprain (Sitler, Ryan, Wheeler, McBride, Arciero, Anderson & Horodyski, 1994; Woods et al., 2003). This increased risk of injury is related to the structure and functional demand of this ligament and the ankle-foot complex during lateral movements.

The ankle joint comprises of the interaction between the tibia and fibula which creates a synovial joint with a deep socket or mortise that contains the talus (Whiting & Zernicke, 1998). In the dorsi-flexed position, the talus fits snugly within the mortise and is considered stable. However, the ankle joint is fairly thin and week in the anterior and posterior directions, and during ankle plantar flexion, this produces a narrower posterior section as the talus rotates into the area between the malleoli, which creates a looser fit and a more unstable joint (Whiting & Zernicke, 1998). This influences the joint laxity, which varies throughout the gait cycle, and changes which ligaments are loaded, as well as the range of the load applied. As the magnitude of plantar flexion increases, so does the instability of the joint and this allows the rearfoot to enter a greater range of inversion (Fujii et al., 2005). This increased inversion places greater loads on to the ATFL whilst the added plantar flexion also contributes independently to an increased strain on this ligament (Colville, Marder, Boyle & Zarins 1990; Ozeki, Kitaoka, Uchiyama, Luo, Kaufman & An, 2006). Further still, the ATFL is the weakest of the stabilising ligaments (Wolfe, Uhl, mattacola, & McCluskey, 2001; Barrett & Bilisko, 1995). The maximum force tolerance is 138 N for the ATFL and 345 N for the CFL (Clanton & Porter, 1997). Therefore, because of these facts and the observation that in a plantar flexed and inverted position the ATFL is relatively taut (Clanton & Porter 1997), this ligament is most commonly damaged (Hockenbury & Summarco, 2001; Colville et al., 1990; Pollard, et al., 2002; Wolfe et al., 2001). Therefore, it is important to understand the risk factors associated with the increased risk of ankle ligament damage, particularly to the ATFL, and how these factors induce changes in foot positions.

Little is known about the specific aetiology of ankle injuries such as sprains (Stacoff et al. 1983; Engebretsen, Myklebust, Holme, Engebretsen, & Bahr, 2008), although it is thought that it is most probably multi-factorial, in which intrinsic and extrinsic risk factors combine to play their part (Willems, Witvouw, De Cock, De Clercq, 2003).

There are many non-modifiable and intrinsic risk factors have been associated with lateral ankle sprains. Ligamentous instability related to previous injury (Tyler, McHugh, Mirabella, Mullanney & Nicholas, 2006; Chomiak, et al., 2000; Ekstrand & Gillquist, 1983b; Woods et al., 2002), anatomical foot and ankle alignment (Cowen, Robinson, Jones, Polly & Berry, 1994), and gait mechanics (Michelson, Hamel, Buczek & Sharkey, 2004) have been implicated as causal factor which can be related to biomechanical changes. Biomechanical gait abnormalities resulting from muscular imbalances, and proprioception linked to the gait mechanics, are amongst the most prevalent factors in the aetiology of ankle sprains (Willems, De Clercq, Delbaere, Vanderstaeten, De Cock, Witvrouw, 2006; Willems, Witvrouw, Verstuyft, Vaes, & De Clercq, 2002). This risk of ankle damage can also be brought about, or accentuated by changes to the boundary conditions such as the footwear and playing surface (Willems, Witvrouw, Delbaere, De Cock, & De Clercq, 2005). This is because shoe and the playing surface can influence the traction provided to the participant (Blazevich, 2004). This can influence the direction and magnitude of the applied force, which determines the level of ankle joint laxity (Fujii, Kitaoka, Luo, Kura, & An, 2005). This can also be accentuated by the speed of the sideways movement (Stacoff, Stuessi & Sonderegger, 1983). This is because both of these factors can lead to increased horizontal forces applied to the ankle joint. Consequently, this encourages greater rearfoot movement, producing greater loading on both the lateral collateral ankle ligaments and musculature, as they try to resist the increasing movements.

2.3.4.3. Metatarsal injury.

The metatarsals are five bones positioned in the forefoot (Figure, 2.3) that act to provide a broad plantar sole in order to distribute the load applied by the body. These structures must be flexible (except the third which is fixed) to aid in balance and the maintenance of the upright posture of the body. The metatarsal also help during the propulsion phase of the gait cycle and the initiation of plantar flexion and phalangeal flexion, where the metatarsals act as a rigid lever to help with forward directed movements (Hamill & Knutzen, 1995).

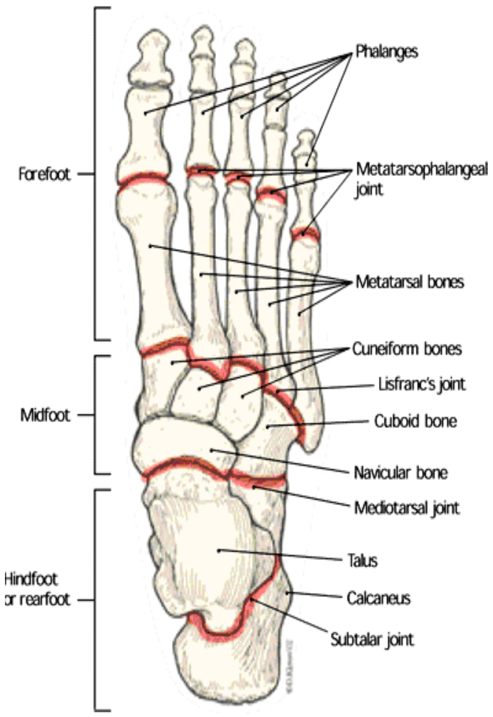


Figure 2.3. Diagram of the foot including the location of the metatarsals and metatarsal joints (Taken from Gore & Spencer, 2009 with permission of the author)

It has been reported that 35% of all metatarsal injuries occur in sports using studded footwear such as soccer (Porter, Duncan & Meyer, 2005). In soccer, repetitive running and turning movements can cause repeated cyclic and impact loading of the metatarsals and this has been linked to deformity, crepitus, swelling, and bruising (Roy, 2006), as well as stress fracture injuries (Frey, 2003; Queen et al., 2008). Stress fracture of the metatarsals in football players has become recognised as a significant injury issue, especially at the higher levels of play (Hussain, 2007; Orendurff, 2008). It has been estimated that soccer has one of the highest frequency of stress fracture compared to other sports. For example, during the soccer World Cup played in the summer of 1994, 38% of the teams had players with stress fractures to the foot (Knapp, Mandelbaum & Garrett, 1998 cited in Queen et al., 2008). A stress or fatigue fracture to the metatarsal is a break that develops in bone after cyclical, submaximal loading (Hockenbury, 1999). According to Wolff's law, bone remodels along lines of stress (Hockenbury, 1999). Bone is constantly being re-absorbed and replaced as the re-absorption of circumferential lamellar bone is accomplished by osteoclasts and replaced with dense osteonal bone by osteoblasts (Hockenbury, 1999). In states of increased physical activity, bone is re-absorbed faster than it is replaced, which results in physical weakening of the bone and the development of micro-fractures (Hockenbury, 1999). With continued physical stress these micro-fractures coalesce to form a complete stress fracture. A "stress reaction" occurs when the micro-fractures are attempting to heal and a complete fracture has not yet developed. Stress fractures are defined as spontaneous fractures of normal bone that result from the summation of stresses, any of which in isolation would be harmless, but become significant when applied together (Rupp & Karageanes, 2008).

To assess the risk of injury such as bone deformity, crepitus, swelling, bruising and metatarsal fracture, high localised pressure across the metatarsal region has been recognised and linked to foot pain and damage, although no definite relationship has been established (Coyles & Lake, 1999).

The high proportion of metatarsal injuries in studded footwear sports such as soccer (Porter et al., 2005) may result from the location of the studs. Lafortune (1998) reported the sensation of increased pain above these traction providing devices. This pain occurred particularly at the first and fifth metatarsals, although no direct link has been demonstrated. However, it may be speculated that the pain resulted from an increased localised pressure and could lead foot damage (Shorten, 1998). This suggests that improper boot design may play an important role in causing excessive loads to be applied to the forefoot and the underlying metatarsal bones (Orendurff, 2008). Therefore, when consideration is given towards the choice of footwear to be worn, soccer players must not only choose the correct boot to avoid ankle ligament injury, they must also appreciate the risk of metatarsal fractures associated with the boot design. The occurrence of stress fractures to the metatarsals has been linked to a hardening of the playing surface (Iwamoto & Takeda, 2003; Queen et al., 2008). This may further increase the loads on the metatarsals, but also interact with the footwear to determine the loading on the structures. Other aetiological factors are also present during metatarsal injury. These include bone strength, bone density and a change in the frequency of loading through an increased running speed and a reduction of time between bouts. These aetiological factors also include the duration of activity, along with a rapid or improper warm-up, overuse, and intense workouts (Rupp & Karageanes, 2008). These may work independently to bring about injury or interact with the surface and footwear condition to heighten the risk further.

Repeated bouts of load applied through the stud on a hard surface may result in a stress fracture occurring. This can be particularly problematic if it occurs at the fifth metatarsal. Injury to this structure is commonly experienced at its proximal base, distal to the fourth-fifth metatarsal base articulation, and usually 1.5 cm distal to the tuberosity (Hockenbury, 1999). The proximal fifth metatarsal has a poor blood supply and is at significant risk of delayed union or non-union. Thus, it is important that the risk of injury to this structure is reduced.

In summary, within sections 2.3.4 injury to the Achilles tendon, ATFL and metatarsals have been highlighted as common injury problems in soccer. Playing surface and footwear are two external and modifiable risk factors that have been frequently highlighted in the literature and are strongly linked to these injuries. Therefore, further investigation is needed into how these factors influence the occurrence of injuries. Likewise, it is important to establish and implement prevention methods in order for these injuries to be reduced. However, before such concepts will be considered, it is important at the stage to review the different biomechanical descriptors that can be used to indicate injury risk as well as research investigations which has found statistical differences in these measurements when risk factors are changed.

2.3.5. Biomechanical descriptors.

In the evaluation of risk factors associated with injury, biomechanical descriptions should be taken to quantify the effect of different risk factors and interventions (Bahr & Holme, 2003; Bahr & Krosshaug, 2005; Meeuwisse, 1994). To understand these biomechanical descriptions, one must start by understanding the structure of the lower extremity and how they become injured.

The frequency of lower extremity injury, particularly those to the ankle and foot, can be traced readily to the foot's complex structure, the need to sustain large weight bearing stresses, and the multiple and somewhat conflicting functions that the foot must perform (Norkin & Levangie, 1992; Whiting & Zernicke, 1998). The ankle/foot complex must be able to provide a stable base to support the body during a variety of weight-bearing postures, whilst also acting as a rigid lever for an effective push-off during gait. Therefore, the stability requirements of the foot-ankle complex can be contrasted by the structure's mobility demands. These stability requirements are necessary for the dampening of the rotations imposed by the more proximal joints of the lower limbs to prevent excessive movements. However, the structure must be flexible enough to absorb the shock of the superimposed body weight as the foot hits the ground, and also permit the foot to conform to the changing and varied terrain on which it is placed (Norkin & Levangie, 1992; Whiting & Zernicke, 1998). Therefore, a compromise is often made regarding the perceived risk of injury resulting from each foot function.

2.3.5.1. Lower extremity loading.

Wear related damage can occur through excessively high friction. However, Radin, Parker, Pugh, Steinberg, Paul & Rose, (1973) wrote that there is very little evidence to suggest that the initiation of lower extremity joint damage is generated from friction between joints as the coefficient of friction within the joint is commonly too low for wear to occur through rubbing. Instead, they suggest that excessive longitudinal forces, resulting from impacts, can predispose the athlete to injury.

Most daily activities involve loading patterns of a rather impulsive nature (Radin et al., 1973), where longitudinal forces of a relatively large magnitude are developed in a relatively short period of time (Nigg, Cole, & Brüggerman, 1995). The term impulsive force encompasses the term impact force, although it is more comprehensive than this force alone (Nigg et al., 1995). However, it is this impact force that receives most attention in research literature.

Impact forces refer to those forces that are generated when two objects collide. Such force is generated during bipedal locomotion when the distal segment of a system, typically the foot, comes into contact with the ground. At this point a state of disequilibrium occurs in the segment, and due to the instantaneous state of stress and strain at the contact point, the impact forces cause the particles in the material at the distal point to move according to the instantaneous stress distribution. This can result in an internal longitudinal stress or shock wave to be generated, which is quite substantial in magnitude (Nigg et al., 1995). These propagate the limb segments at a speed dependent upon the elastic modulus and density of the material it passes (Nigg et al., 1995).

The magnitude of impact force determines the size of the initial stress waves that travel through the body to the head. The rate at which the load is applied also determines the rate at which this force travels through the body. The larger the magnitude and rate of loading of the impact force, the greater and more rapid the shock wave that travels through the lower extremity. At a discontinuity, such as a joint, the wave is transmitted proximally through the joint but is also reflected back towards the source. These reflected stress waves travel back in the direction at which they were originally sent and interfere with the waves that are continually propagating from the area of contact (Nigg et al., 1995). The finite speed of propagation and superposition of wave upon wave creates a non-uniform distribution of stress within the material at the distal end of the segment system. The size of this stress wave is therefore dependent on the frequency of the internal force, which had determined the original wave length within the material (Nigg et al., 1995). This stress can cause micro-trauma in the joint and surrounding structures, which the biological system uses to adapt and remodel to provide strong (Radin, Orr, Kelman, Paul & Rose, 1982) and healthy tissue (Derrick & Mercer, 2004). However, this micro-trauma may result in injury. The adaptation is therefore important, but is dependent upon four factors. The first two factors are the magnitude of the internal load and the tensile limit of a structure. The ability of the human body to adequately absorb high-impact forces is essential in the prevention of injury (Hargrave, Carcia, Gansneder, & Schultz, 2003; Radin et al., 1973). Therefore, if the load is greater than the tensile limit, damage will occur. The third factor is the loading characteristics of the force (e.g. impact and cyclic) which determines the rate and frequency of loading application. The fourth is the time given for the structure to recover. If all of these factors are optimal, moderate intermittent loading of bones, ligaments, cartilage and tendons will enable the structures to develop in order to attenuate and withstand larger magnitudes of force. However, the integrity of a joint complex may be reduced, if the frequency, magnitude and time frame between the repeated sub-maximal loading bouts do not allow adequate rest for the remodelling process to be completed. In this case the micro-trauma damage will be sufficient for injury to result.

It is difficult to measure the shock waves associated with causing injury using in-vivo methods (Bobbert, Yeadon & Nigg, 1992). Instead, reliable information can be obtained by measuring the magnitude of ground reaction forces (GRF) (Bobbert, et al., 1992). Ground reaction force is a measurement used to assess Sir Isaac Newton's third law of motion which stated that "for every action there is an equal and opposite re-action". Ground reaction force is the combination of three orthogonal force vectors (vertical, medio-lateral and posterior-anterior) (Figure 2.4), that are applied by the ground but which are opposite in direction and equal in magnitude to that which the body applies to the surface during any weight bearing activity. Vertical ground reaction forces (VGRF) are commonly measured as an estimation of the size of the impact forces and the magnitude of the internal shock wave.

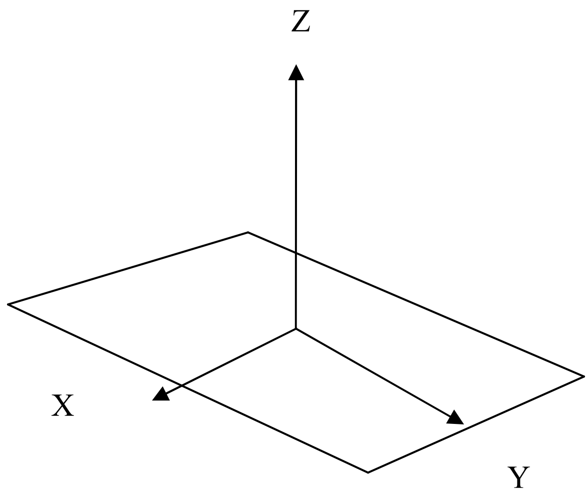


Figure 2.4. Representation of the three orthogonal force vectors that contribute towards ground reaction force (GRF). Where X represents the medio-lateral force, Y the anterior-posterior force and Z the vertical force. (Adapted from Kwon3d, 2009).

Vertical ground reaction force represents the change in vertical acceleration of the centre of mass (COM) of the body. The magnitude of the COM acceleration is influenced by different segmental accelerations in which each can move at different rates throughout the gait cycle. The changes in velocity can be seen in the VGRF-time history curve as distinct force peaks (Nigg et al., 1995). The shape and magnitude of these force-time curves can be influenced by the running style of an individual. Participants can be categorised as having a heel-toe, mid-foot or forefoot running style force-time profiles. Most athletes possess a heel-toe running style (Nigg et al., 1995). The VGRF-time curve of these athletes is typically double peaked (Figure 2.5). The first peak represents the "passive" phase whereas the second peak represents the "active" phase. In the passive phase (otherwise known as impact phase or heel strike), the peak reflects the initial impact between the body and the ground and is determined by conditions evident at impact (Bobbert et al., 1992). This is identified as the maximal amplitude of the vertical force during the first 50 ms (Bobbert et al., 1992; Nigg et al., 1995) and is used as an estimate of the peak impact force and internal shock wave. This is commonly scaled with body mass (Cavanagh & Lafortune, 1980; Munro, Miller, & Fuglevand, 1987), and is reported to be between 2 to 5 times the body weight of the performer (Cavanagh & Lafortune, 1980; Frederick, Hagy, Mann, 1981; Nilsson & Thorstensson, 1989). Many factors influence the acceleration of the COM and thus the magnitude of the VGRF at impact. These include gender, age, body mass, running speed (Mercer, Vance, Hreljac & Hamill, 2002; Nilsson & Thorstensson, 1989; Perry, & Lafortune, 1995), stride length (Martin & Marsh, 1992; Nilsson & Thorstensson, 1989), muscular activation (Ferris, Liang & Farley, 1999; Ferris, Louie, & Farley, 1998; Gerritsen et al., 1995), skeletal alignment (Stacoff et al., 1988) and the wobbling mass of the participant (Pain & Challis, 2004). On the other hand, the active phase (also referred to as propulsive or toe-off phases) reflects the propulsive forces applied by muscular contractions of musculo-skeletal system to move the COM during this phase.

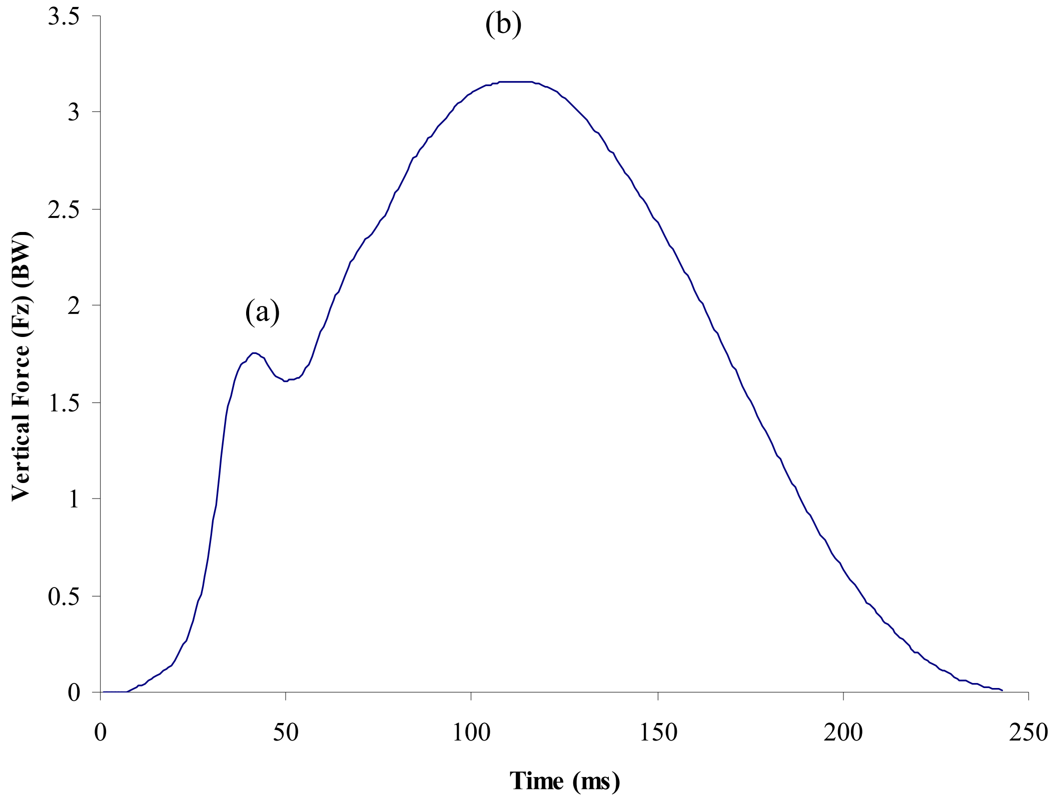


Figure 2.5. A typical vertical force-time history evident during heel-toe running, where peak impact (a) and peak propulsive (b) forces are shown

Many researchers have associated high VGRF at impact with many different sports injuries (Cavanagh & Lafortune, 1980; Clement, Taunton, Smart, & McNicol, 1981; Munro et al., 1987). Further still, in an attempt to ascertain a relationship between VGRF at impact with injury, animal studies have been performed to directly compare the effect of repeated impact loading on different structures of the animal's body. These studies have shown that abnormal stress waves resulting from impulsive loading can cause degeneration to bone and to articular cartilage which can lead to osteoarthritis (Day, Ding, van der Linden, Hvid, Sumner, & Weinans, 2001; Radin & Paul, 1971; Radin & Rose, 1986). Likewise, bone ligament and other biological tissue exhibit viscoelastic behaviour with their load-extension being time-dependent (Noyles, Delucas & Torvik, 1974). As such the mechanical properties of tissue are dependent upon the rate of loading, the rate of elongation or both (Noyles et al., 1974). Therefore, a high rate of impact loading has been associated with increased prevalence of injury (Ewers, Jayaraman, Banglmaier, & Haut, 2000; Lees & McCullagh, 1984). However, despite the evidence linking the magnitude and loading rate of impact forces with injury, Nigg et al. (1995) questioned the relevance of previous findings. The evidence were either derived from animal studies or are circumstantial in nature, where the descriptions of the aetiological factors that are associated with injuries are often based on expert opinion, or evolve from descriptions of injury observations that fail to utilise a control group (Messier & Pittala, 1988). Instead, to establish a relationship between external kinetic measurements and injured and non-injured participants, prospective studies are required (Bobbert et al., 1992).

Hreljac (2004) summarised a collection of research investigations and described that runners who exhibit relatively large VGRF at impact and maximal vertical loading rate while running, were at an increased risk of overuse injury. Hreljac, Marshall & Hume (2000) also found that participants who were injury free experienced significantly lower VGRF at impact compared to those athletes that experienced injury to the lower extremity (injury group= 2.4 BW [S.D. 0.41], injury free = 2.13 BW [S.D. 0.42]). Hreljac et al. (2000) also reported significantly greater risk of injury in participants with a significantly greater peak rate of loading (injury group = 93.1 BW.s^{-1} [S.D. 23.8], injury free group = 76.6 BW.s^{-1}[S.D. 19.5]). Ferber McClay-Davis, Hamil, Pollard & McKeown (2002) showed that participants who have had a previous lower extremity stress fracture exhibited 36% greater peak VGRF at impact, 32% greater instantaneous loading rate, and 34% greater average loading rate, compared to a non-injured control group. Radin, Yang, Riegger, Kish, and O'Connor (1991) also found that human participants who exhibited knee pain showed differences during the first few milliseconds of heel strike, where a significantly greater VGRF at impact was observed, with a faster downwards velocity of the ankle and angular velocity of the shank immediately prior to heel strike in participants with an injury. The follow through of the leg was also said to be more violent with larger peak axial and angular accelerations, which matched a more rapid rise in the VGRF.

In contrast to the previous evidence, Bahlsen (1989 cited in Nigg et al., 1995) showed that during a six month period, runners who had high VGRF at impact at the beginning of the study suffered no significant increase in short term injuries compared to those who had lower forces. Likewise, Stefanyshyn, Stergiou, Lun & Meeuwissse (2001) measured the VGRF at impact and impact loading rate in 143 runners prior to a 6 month running season. The participants were grouped into the bottom (25%) middle (50%) and top (25%) for each measured parameter (i.e. those that had the lowest, middle and highest impact force magnitudes) and the injury rates were compared. They concluded that their results were in contrast to the common belief that high VGRF at impact and impact loading rate were related to running injury, observing no significant differences between the groups. Luethi, Frederick, Hawes & Nigg (1986) also compared VGRF in participants who showed pain to those who did not when performing a lateral movement. These authors found that foot pain during this movement was not associated with the magnitude of vertical or horizontal GRF during these movements.

In light of the evidence from prospective studies, a clear association between the magnitude or rate of loading of the VGRF and risk of injury remains questionable. Consequently, a direct cause-effect relationship remains elusive (Stiles & Dixon, 2007). The difficulty in establishing a direct link between the magnitude of VGRF at impact and injury, may relate to the fact that not all participants with naturally high impact forces will sustain injuries. Protection from high shock waves may be provided to some participants through exercise induced hypertrophy to the structures of the lower leg, or through genetic characteristics that determine the condition and quantity/thickness of the bone, cartilage, ligaments and adipose layer. These factors influence the transition of the force through the lower extremity. This is because the material with the greatest visco-elastic properties can absorb and dissipate the energy of a stress wave as heat within the material more efficiently and attenuate the amplitude of the wave as it passes through the material (Nigg et al., 1995; Radin & Paul, 1971; Voloshin & Wosk, 1982. Voloshin, Wosk & Brull, 1981). In particular, despite the high visco-elastic properties, peak dynamic force attenuation has been shown not to be the function of synovial fluid and articular cartilage, but of bone and soft tissues (Radin & Paul, 1971). These structures are not as elastic as cartilage and synovial fluid but exist in large enough quantities to provide sufficient deformations to act as efficient shock absorbers (Radin & Paul, 1970). These factors increase the length of time taken for the force to pass through the material, and thus reduce the shock wave experienced by the athlete. On the other hand, participants with poor conditioning may be susceptible to injuries even though they possess smaller impact force magnitudes.

Methodological limitations may go some way to explain the ambiguity regarding an association between the magnitude of VGRF or loading rate and injury. For example, when the previous prospective studies were performed, not all factors that increase a participant’s risk of injury were accounted for. These may include previous injuries, training pattern characteristics and magnitudes, and/or anthropometric, biological or biomechanical (kinematic) characteristics. Consideration of these factors may help explain the conflicting evidence provided in the aforementioned prospective studies. However, despite the lack of a direct relationship, a relative change in peak impact forces or rate of loading occurring for the same person under different conditions would indicate a relative change in injury risk. Thus, the measurement of impact forces and loading rates is still warranted as an indicator of injury risk change.

Another possible reason for the failure to establish a direct link between the magnitude of the VGRF at impact and injury may be due to the premise behind the measurement of peak VGRF at impact. The theory suggests that GRF represent the magnitude of the internal shock waves that propagate directly up the lower extremity to the head through the heel, stressing the lower limb structures as the wave progresses through the body (Clinghan, Arnold, Drew, Cochrane & Abboud, 2008; Voloshin & Wosk, 1982). However, Shorten (2002) described that the reaction forces acting on the runner are not applied at a single point. During the impact phase of 0-50ms, the forces are not only focused upon the heel but are also distributed across the mid and fore-foot. Therefore, the resultant impact force does not accurately represent the forces that transcend the lower extremity vertically through the heel (Figure 2.6). Heel forces are therefore more indicative of the strength of the shock waves which perpetuate damage of the musculoskeletal structures surrounding the foot and ankle (Clinghan et al., 2008). Further still, it could be argued that understanding localised heel forces are more important than the measurement of VGRF at impact when trying to understand the causes of impact related injury. Indeed, the former may offer more sensitivity than GRF to detect differences in risk factors such as surface cushioning, which has been linked to injury (Dixon & Stiles, 2003).

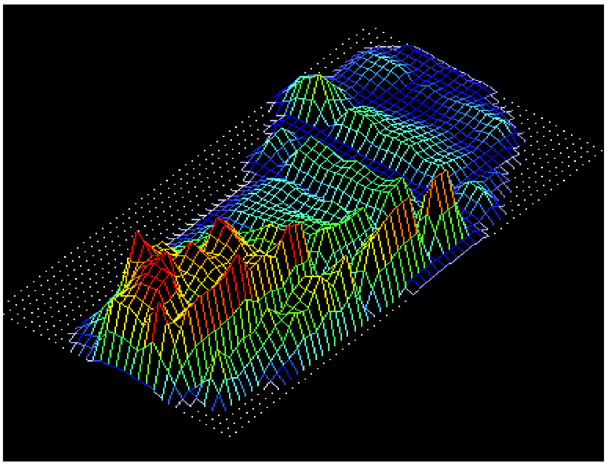


Figure 2.6. A three dimensional image showing varying plantar pressure distribution during first 50 ms of heel contact in a participant performing a running task

In kinetic gait analysis, the magnitude of local pressure data is important (Rosenbaum & Becker, 1997). Both force and pressure measurements detail the loading of the foot. However, despite this association, they are essentially describing two independent factors. The quantification of force describes the interaction between two bodies and the environment. This is measured in Newtons and can be defined as the force necessary to accelerate the mass of 1kg by 1 $m.s^{-2}$. Pressure on the other hand is the measure or analysis of the distribution of force across a surface area and is reported as force per area unit ($N.m^{-2}$ or Pascal where $1N.m^{-2}$ = 1 Pascal) (Rosenbaum & Becker, 1997).

Plantar forces and pressure data have been used to understand possible differences between patients with and without various pathological ailments across the forefoot, and provide important information regarding the loading of the plantar tissues inside the shoe. For example, propulsive forces are created when the body is moving out of the stance phase of the gait cycle and are related to the movement speed of the athlete. To increase the movement speed, greater propulsive force is required. However, propulsive forces also represent the loading that occurs across the toes and metatarsals, and these push off forces have also been associated with injury (Cavanagh & Lafortune, 1980). However, there are many structures located in the forefoot, each of which can become injured. The magnitude of the propulsive force can be used to indicate the magnitude of the load across the forefoot, although it cannot detail the loading of different areas as it is only one force vector. Instead, using pressure technology, the foot can be divided into key areas and a measurement can be made regarding the localised forces. This has been used to determine the localised forces at the toes and metatarsal heads, and differences have been shown between patients with and without neuropathic ulceration (Ctercteko, Dhanendran, Hutton, & Le Quesne 1981; Patel, & Wieman, 1994). However, although localised forces provide the force across an area, it does not indicate the force per unit area, which may be more relevant when trying to identify the loading on a specific structure, such as the metatarsal. Areas of high pressure have been associated with increased pain across the metatarsal region in soccer, particularly above the studs. Therefore, it is possible to calculate loading under the studs of different soccer boots (Coyles & Lake, 1999; Lafortune, 1998). The same could be said for load experienced at the medial and lateral locations of the heel, where the studs are positioned.

Kinetic measurements techniques.

To assess changes in GRF measurements, a method of data collection is required. Traditionally, force plates are used to collect GRF data. Force plates use three dimensional force transducers placed in each corner of a force plate, to measure the three components of GRF, along with centre of pressure and free rotational moment (Rosenbaum & Becker, 1997). The transducers exert a small force which is sent to an amplifier to be converted into a GRF magnitude (Rosenbaum & Becker, 1997). The technical specifications of gait analysis systems have been summarized by Brand (1992, in Rosenbaum & Becker, 1997), who stated that the system should be accurate and reproducible, stable over time, not interfere with the motion of the participant, and cost effective. Force plates meet such requirements, although their main advantage is their ability to record the GRF in all directions and at a high frequency (a typical collection frequency of GRF data is 500-960 Hz), which enables a high level of detail to be maintained during high speed movements (Rosenbaum & Becker, 1997). In contrast, one limitation of traditional force plate technology is that it only measures the resultant vertical force vector. Therefore, the disadvantage of this technology is that the systems do not provide insight into the distribution of the load over the plantar surface of the foot. This causes the data provided to have limited relevance to the anatomy or pathology of the foot (Cavanagh & Rodgers, 1985 in Rosenbaum & Becker, 1997).

The use of GRF to distinguish differences between variables is dependent upon the stability of the data provided, which is influenced by the variability of the same performer and equipment over repeated trials. Runners are biological organisms with anatomical and functional differences. As such, there is some variability between performers and within the same performer over repeated trials (Bates, Osternig, Sawhill, & James, 1983). Bates, Osternig, Mason, and James, (1979) reported that when three consecutive footfalls were compared for a number of selected parameters, no significant differences were shown. However, between trial variability was observed and Bates et al. (1979) suggested that the magnitude of this variability would influence the resulting mean from the multiple trials. Therefore, this would need to be evaluated if subtle differences are to be detected between conditions such as footwear. Bates et al. (1983) investigated the number of trials necessary for stable data. They found that the mean score from a minimum of eight trials was necessary to obtain stable subject-condition values of GRF when comparing subjects, shoes or the various curve parameters, and the mean from ten trials are necessary in order to obtain results at the 95% confidence interval. This, they stated, ensures that the cumulative mean value falls within a criterion value for the general population for both medio-lateral and vertical force estimates. However, DeVita and Bates (1988) suggested that based upon independent parameter variability and minimum sample size evaluations, 25 trials are necessary for accurate GRF data which can detect subtle real differences between shoe conditions. Because of the contrasting evidence regarding data stability and the number of trials needed, consideration of the data stability should be given prior to biomechanical research.

Due to the increasing appreciation of the limitations of GRF measurements, clinicians and researchers have increasingly utilised pressure technology (Woodburn & Halliwell, 1996). Pressure measuring devices generally offer greater flexibility than traditional force plates (Martinez-Nova, Cuevas-García, Pascual-Huerta, & Sánchez-Rodríguez, 2007). These permit the most important interface, which is located between the foot and shoe, to be analysed (Cavanagh, Hewitt, & Perry, 1992). Using this technology, conclusions have been made based on subtle changes in vertical force at local regions and the specific load characteristics at key areas across the foot (Martinez-Nova et al., 2007). This method can also be used away from the constraints of the biomechanics laboratory, allowing for a more versatile and robust measurement (Cavanagh et al., 1992). It also removes the inherent targeting problems associated with force plates (Martinez-Nova et al., 2007; Santos et al., 2001) and providing data for multiple steps during one trial, which is not permitted with the use of one force plate. These benefits are all provided whilst offering minimal interference with the foot function (Martinez-Nova et al., 2007).

Many manufacturers have developed different versions of the pressure insole principle, each in slightly different forms and with varying technology (Novel Pedar system; Footscan system; F-scan system). These systems have been used in both biomechanical (Dixon et al., 1998; Stiles & Dixon, 2006) and medical situations (Hosein & Lord, 2000; Lord & Hosein, 2000). Pressure insoles use technology that calculates the pressure from the normal force acting perpendicular to sensors typically constructed from piezoelectric crystals. With each force application, the individual insole sensors emit an electrical impulse that has a known voltage to force ratio. The magnitude of the force for one or multiple sensors can be used to provide total force within a given anatomical area. Pressure masks can also be set to provide pressure measurements within key areas of the plantar foot by dividing the magnitude of force by the area of the mask (Figure 2.7). The sum of the individual sensors can also be used to calculate the resultant force. However, a limitation of pressure insoles is that they do not typically measure shear forces acting horizontally.

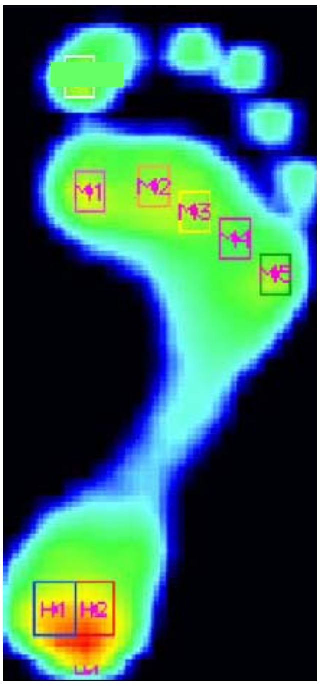


Figure 2.7. Example of plantar foot pressure image and pressure masks used to calculate plantar pressures at places of interest. M1-5 represents the masks placed upon the metatarsal heads, and H1 and H2 define the medial and lateral heel masks respectively.

Pressure insole data can be influenced by the reliability of repeated measurements, which along with the accuracy of pressure insoles are important considerations for pressure insole use. Reliability refers to the ability of the pressure insole to measure repeated measurements consistently. This is influenced by variations in movement patterns of participant. It can also be influenced by the consistency of the pressure measuring device used. As the consistency of the measurements increases, there is less variation in the data, so is more stable. This variability contributes towards whether the mean of the measurements are representative of those experienced by the individual. Therefore, the reliability indicates the consistency or stability of both the data measured by the insole and the participant. This reliability can depend upon the number of trials used, which tends to increase as the number of repeated measures goes up. Thus, it can be used to indicate the number of trials necessary for a stable representation of the biomechanical variables experienced. This can depend upon the manufacture and model of the insole.

Boyd, Bontrager, Mulroy & Perry (1997) used an Intraclass Correlation Coefficient (ICC) to assess the reliability of the Novel insole system during a walking task. They found that with the use of 10 trials the ICC of the peak pressure magnitude at the heel and lateral arch were excellent (0.82 and 0.89), although the medial arch showed poor reliability ($r < 0.50$). Boyd et al. (1997) also showed that the pressure measurements at all metatarsal areas demonstrated good reliability as did the pressure occurring at the small toes, although pressures measured at the large toe showed poor reliability.

To measure pressure insole accuracy, the force plate is commonly used as the gold standard. Using this technique, the accuracy of the Novel Pedar insole was assessed by Boyd et al. (1997) by comparing the vertical forces measured by the insoles with those attained force plate. During a barefooted walking task, significantly greater forces were recorded with the force plate compared to the pressure insole for the first peak but not the second. Barnett, Cunningham & West (2001) reported an overall mean underestimation of peak force was an average of 13.4% with the insole. Because of these findings, Barnett et al., (2001) recommended that when using pressure analysis for clinical applications, the underestimation should be taken into consideration. To explain this underestimation, Barnett et al. (2001) described that the insoles measure the normal force acting on the sensor which is not necessarily in a vertical position. Thus insole data is not always the same as vertical force measured by the force plate. Because of this, the magnitude of the underestimation is associated with the position of the foot at contact, influencing the angle of the sensors relative to the direction of the vertical force. The underestimation in force magnitude may also be somewhat dependant upon the tasks performed, the period of the gait cycle, and the gait style of the individual. Further still, there is "dead space" located between each sensor and this can result in some of the resultant force not being measured by the insole (Barnett et al., 2001; Kaplen & Sietz, 1994). Finally, repeated loading over time and insole temperature has been shown to increase the underestimation of the vertical force and pressure magnitudes, influencing the accuracy and the reliability of data provided by repeated trials (Cavanagh et al., 1992; Hughes, Pratt, Linge, Clark, & Klenerman, 1991; Hurkmans, Bussmann, Selles, Hormans, Benda, Stam, Varhaar, 2006).

An alternative pressure insole system used during running related research (Dixon et al., 2008; Stiles & Dixon 2007) is the RSscan pressure insole (Footscan, RSscan International, Belguim). The insole is 0.7mm thick and is made of peizo-electric sensors that are 5-7mm in size and has a resolution of 3 sensors per cm^2, and collects data at a frequency of 500 Hz. However, to the authors' knowledge there is no literature available assessing the reliability and accuracy of this insole; this is analysed and discussed in Appendix B and C.

Using the combination of force plate and pressure insole technology to attain the various parameters of lower extremity loading, it is possible to assess the risk factors such as playing surfaces and footwear associated with overuse injury to the Achilles tendon and metatarsals.

2.3.5.2. Kinematic descriptors.

As discussed previously, the peak magnitude and loading rates of VGRF, and regional forces and pressures, influence the risk of injury during athletic performance. However, it has been suggested that various kinematic characteristics are also related to the risk of sustaining athletic injury. The ability for the most distal structures of the human body to absorb high-impact forces and to lower the loading is essential for injury prevention (Hargrave, et al., 2003). Therefore, one way in which kinematic variations can influence injury risk is through the determination of the impact force magnitude. The primary method by which running participants can manipulate the magnitude of force is through pronation of the subtalar (talocalcaneal joint) joint (Whiting & Zernicke, 1998), where an increase in the initial pronation can cause a decrease in impact forces and vice versa (Stacoff, Denoth, Kaelin, & Stuessi, 1988). To understand how pronation influences the level of force experienced by the body, it is important to understand the structure of the subtalar joint (talocalcaneal joint), which combines with the ankle joint (talocural joint) and the midtarsal joints within the foot-ankle complex.

Subtalar and talocural joint.

The subtalar joint is the main joint in the rearfoot (Hennig, 2002). It is a composite joint formed by three separate plane articulations between the talus (superiorly) and the calcaneus (inferiorly) (Figure, 2.8). The articulation of the calcaneus with the talus is essential for proper function of the foot-ankle complex during load bearing (Whiting & Zernicke, 1998). It has to be flexible enough to attenuate force whilst also being rigid to dampen the rotational forces imposed by the body weight during ground contact, thus preventing excessive movements.

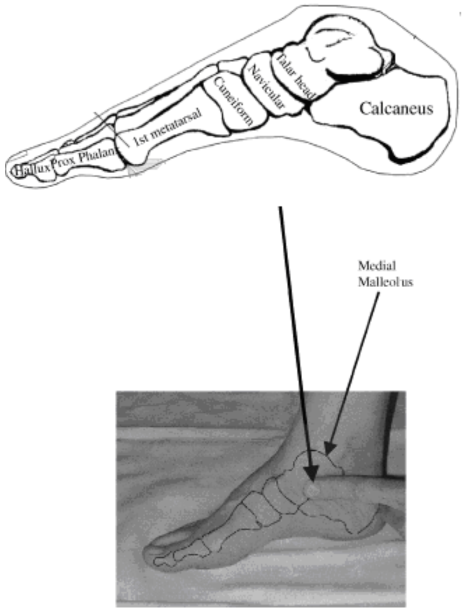


Figure 2.8. Diagram showing the articulation between the talus and calcaneus to form the subtalar joint (Taken from Rothbart, 2002, with permission from Journal of Bodywork and Movement).

Subtalar joint pronation is a triplanar movement which produces the combined movements of dorsi-flexion and abduction of the talus, and calcaneal eversion (Hennig, 2002). The subtalar joint also influences the magnitude of supination which is the opposite movement of pronation (plantar flexion, adduction and inversion), and is considered to be an important measurement during lateral movements (Hennig, 2002) (Figure 2.9). The movements of the subtalar joint cannot occur without one another, although the ratio of the different components is not equal. For every 4° of calcaneal eversion, one should see 4° of adduction of the talus (4° of medial rotation of the tibia) and 1° of plantar flexion of the talus (Whiting & Zernicke, 1998). The range of the eversion/inversion, abduction/adduction, and dorsi-flexion/plantar flexion movement for the average subtalar joint axis is therefore 4:4:1 (Norkin & Levangie, 1992; Whiting & Zernicke, 1998). A far greater contribution towards ankle dorsi-flexion and plantar flexion movement is given by the talocural joint which will be discussed later in this chapter.

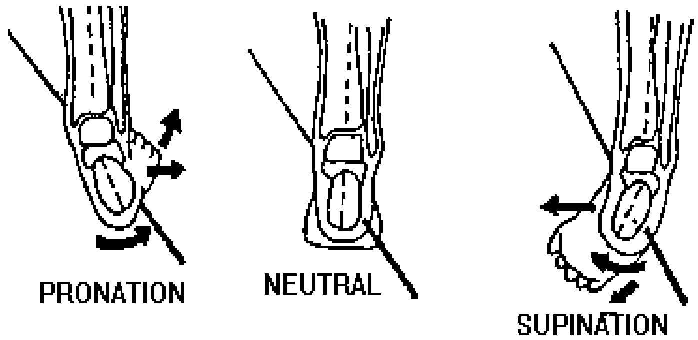


Figure 2.9. Example of subtalar joint supination and pronation of the right foot (Taken from Thompson, 2001)

In providing increased pronation, the mid-tarsal joint is unlocked which depresses the medial longitudinal arch, allowing the foot to become flexible and absorb shock during weight bearing (Hargrave, et al., 2003). Along with the unlocking of the mid-tarsal joints, the force attenuation associated with pronation also relates to the foot position at initial contact and the range of movement provided throughout the rest of the contact phase. The degree of the total pronation influences the magnitude of the force experienced by the athlete through the manipulation of the size of the lever arm produced by the foot. During the touchdown phase of gait, athletes tend to land in a slightly supinated position. As such, the line of action of the impact force does not pass through the subtalar joint during the initial pronation period. This is represented by the fact that the centre of pressure is located laterally from the subtalar joint (Figure 2.10) (Stacoff et al., 1988). The further towards the lateral side of the heel the centre of pressure is initially positioned, the greater the distance this force has to travel to cross the subtalar joint, thus creating a larger lever arm length (a) (Figure 2.10). The degree of lateral heel contact also influences the length of time the foot is decelerated (de Wit, de Clerq & Lenoir, 1995; Stacoff et al., 1988). The more lateral the position of the GRF, the greater the moment about the joint. This causes the foot to enter a fast initial pronation to make the deceleration more rapid (Denoth, 1986; Stacoff et al., 1988). Therefore, the GRF is reduced, as there is less acceleration of each kilogram of body mass for the ground to resist/decelerate.

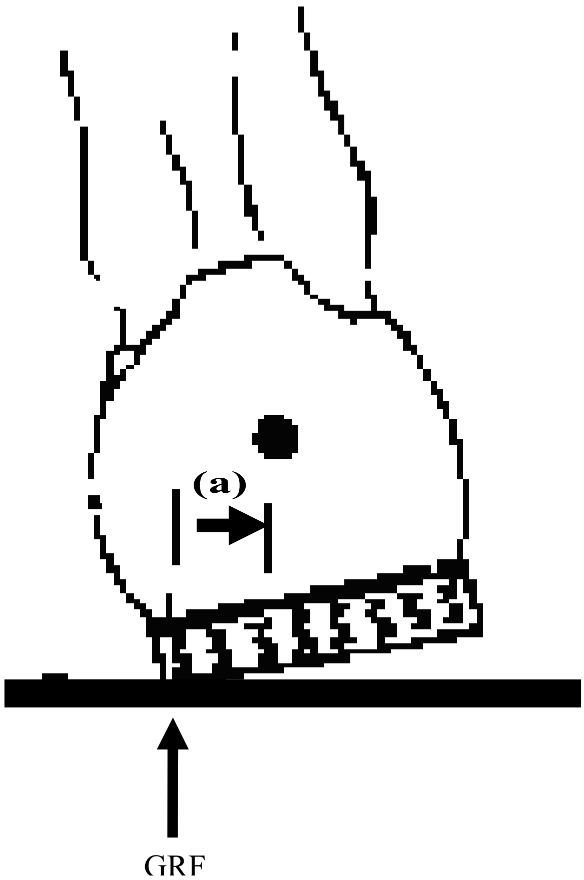


Figure 2.10. Position of GRF force vector at initial contact of the left foot with the ground (Taken from Stacoff et al., 1988)

Although the increased lever arm length acts as a mechanism by which peak GRF is lowered, the greater distance and deceleration time necessary to provide such reductions, also combines to influence the peak magnitude of a joint moment (Hennig, 2002 Stacoff et al., 1988). With large pronation movements, large levers are experienced and muscular and ligamentous loading on the medial side may experience a sudden load increase to control the movement, which may cause an increased risk of injury of the ankle musculature and ligaments during running (Stacoff et al., 1988). This increased initial pronation, can also increase the load on the lateral aspect of the foot. This is because the greater deceleration causes an increased eccentric loading of the inverting muscles (De Wit et al., 1995). Further still, Stacoff et al. (1988) explained that despite large lever arms (2 cm) being beneficial at reducing impact force, runners with an inherent varus alignment (the distal segment deviates medially with respect to the proximal segment), contact the ground in an excessively supinated position. As such, the centre of pressure is very close to the lateral side of the foot and must therefore undergo a greater range of motion in order to achieve flat contact with the ground, which is termed functional over-pronation. This causes very large levers (3-4 cm) to be evident and this reduces the resultant angular pronation velocity. This is because with larger levers the force maxima are reached later in time (Stacoff et al., 1988). As a consequence, the muscles can more easily stabilise the movements about the joint, thereby reducing the angular velocity (Stacoff et al., 1988). This can cause a participant to experience a similar or even greater peak GRF, than those participants with lower magnitudes of pronation. Excessive pronation is also linked to numerous lower-limb ailments (Stacoff et al., 1988). These can include tendonous and ligamentous problems at the shank and knee (Clement, et al., 1981), as well as inflammation at the insertion points for several muscles including the Achilles tendon (Nigg, 1986). Also, as the foot and knee are mechanically linked, the magnitude of pronation, specifically the internal rotation of the foot, has been shown to influence the magnitudes of tibial internal rotation (TIR) (Lundberg, Svensson, Bylund & Selvik 1989). Therefore, when pronation is excessive, extreme internal knee rotation occurs, resulting in increased stress and strain to the ligaments of the knee and the structure of the shank (Lundberg et al., 1989), which increases the risk of injury to these structures.

To measure the risk of injury, the range of motion at the subtalar joint during subtalar supination and pronation is needed. This can be difficult to assess due to the tri-planar movement at this joint, and because the range of movement varies with the inclination of the subtalar axis (Whiting & Zernicke, 1998). Therefore, the measurement of subtalar joint pronation is typically performed by viewing calcaneal eversion (Clarke, Frederick, & Hamill, 1983; Shorten, 2000; Whiting & Zernicke, 1998). This is because the magnitude of calcaneal eversion is dependant on the amount of adduction and plantar flexion (Clarke et al., 1983; Norkin & Levangie, 1992; Whiting & Zernicke, 1998). Rearfoot eversion is also relatively easy to measure whilst weight-bearing and non-weight-bearing.

The pattern of eversion movement during locomotion has been used extensively to evaluate various risk factors. During normal running, the foot contacts the ground beneath the COM of the body, in order to facilitate balance. The foot naturally makes initial ground contact in a slightly inverted position, where the rearfoot typically lands at a position of 0 to 10 ˚ relative to the lower leg (Shorten, 2000). Eversion then occurs as the foot rotates to make flat contact with the ground, reaching a maximum position of eversion about midway through the ground contact period. In normal runners eversion ranges between 8 to 15˚, and the total range of eversion from first contact to mid-stance is typically 10-20˚, but may exceed 25˚ in some subjects (Shorten, 2000; Whiting & Zernicke, 1998) (Figure 2.11).

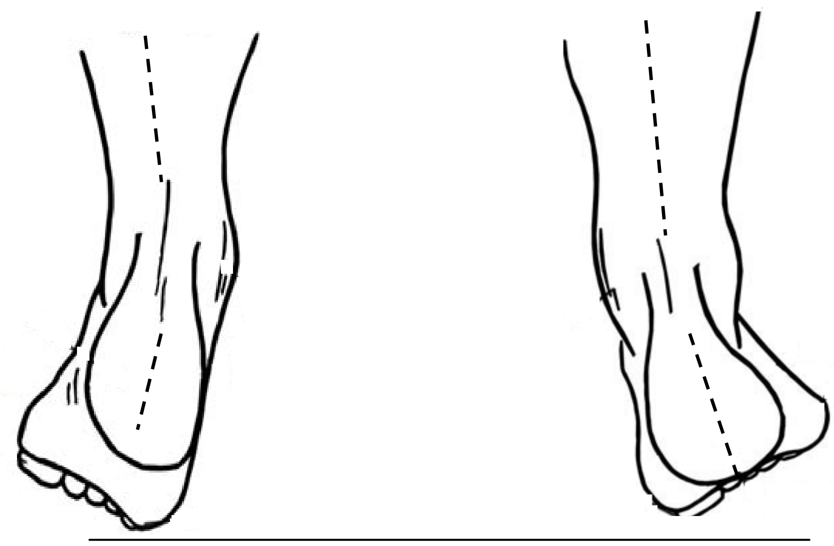


Figure 2.11. Images showing rearfoot inversion of the right rearfoot (left) and eversion of the right foot (right) of the left foot (Adapted from Dobrowolski, 2008 with permission of the author)

To support the theory that increased pronation can cause a lowering of impact forces, Stacoff et al. (1988) modelled the effects of eversion on impact forces using the rearfoot eversion angle as a measure of pronation with a two-dimensional frontal plane model. It was found that with less rearfoot eversion of the subtalar joint, a reduced lever arm was experienced, and consequently, higher impact peak forces were reached at an earlier time. Perry and Lafortune (1995) also investigated the role of pronation in impact force attenuation, but used the biomechanical assessment of a group of participants. Again using rearfoot eversion as a measure of pronation, the authors found that when participants ran in varus footwear that purposely reduced pronation, higher peak impact forces were evident compared to valgus and normal footwear that caused greater levels of pronation. However, when the impact forces were compared between the footwear that exaggerated pronation (valgus footwear), no reduction in peak force was observed compared to normal running shoes, suggesting that there is an optimal level of pronation to lower impact forces. Beyond this level no additional attenuation is provided. Perry and Lafortune (1995) also found that when the same footwear was tested during walking, the GRF variables were not significantly different for any of the footwear types, suggesting that the type of locomotion was important for pronation to attenuate impact forces.

Although force reduction in running may be provided through pronating the foot in sports such as soccer, movements do not always occur in a straight line. Instead, cutting and turning movements are common and these are described as lateral movements. In these tasks, inversion is the main movement of the rearfoot. Luethi et al. (1986) described that lateral movements are very complex and that great variations are expected between and within the participant group. Stacoff et al. (1983) found that during lateral movements, the average range of total movement (peak inversion to peak eversion) for one subject was 20° when wearing a variety of shoes and the range between subjects was 43°. This showed that the range of variability for all subjects during lateral movements equalled about twice the variability of one, highlighting the individual nature of turning movements. For comparison, Stacoff et al, (1983) highlighted that the typical range of eversion when running is approximately half of the range of inversion during sideways movement. These discrepancies may be the result of different anatomical constraints between eversion and inversion, the type of exercise routine performed, and ability for the movement to be standardised (Stacoff et al., 1983). However, this average range was calculated with only 5 trials. Therefore, if more trials were used this average may be reduced. Still, consideration must be given towards the effect that these factors may have on the values attained when risk factors are assessed.

The benefit of greater inversion movement is that it provides a natural mechanism by which impact force attenuation is provided during lateral movements (Dayakidis & Boudolos, 2006). Similarly, additional plantar flexion (which will be discussed in detail later) can also serve as a method to reduce impact force during lateral movements (Brizula, LLana, Ferrandis & Garcia-Bellenguer, 1997). However, during these tasks, a greater range of plantar flexion can allow the foot to enter greater ranges of inversion (Fujii et al., 2005), and as discussed in section 2.3.3, this greater movement increases the risk of inversion injury.

Luethi et al. (1986) compared two groups, one which exhibited pain during turning movements and another which did not. The authors reported similar vertical and horizontal forces between the groups, but observed increased inversion magnitudes in those participants classified as experiencing pain. Because of the association between kinematic changes and impact force regulation, this observation may suggest that some participants in the pain group may have increased their rearfoot inversion in order to obtain manageable impact forces. The pain experienced may therefore be related to increased loads placed on the lateral ankle when entering greater ranges of inversion movement.

Stacoff et al. (1983) found that during lateral movements, maximum inversion occurred within the first 50 ms after touch-down, which is approximately the same time span that the maximum magnitude of pronation occurs during running. This may provide sufficient time for voluntary muscular control of the ankle during supination, in order to protect the ligaments of the joint. The muscles that provide protection during lateral movements include the peroneal muscles (Longus and Brevis) which cause eversion and plantar flexion of the ankle, whilst also resisting the inversion motion. These muscles are activated prior to foot contact during tasks such as jogging (McLoda, Hansen, & Birrer, 2004). However, to increase the inversion magnitude so that to attenuate greater forces, the peroneal muscles must exert less force and thus the lateral ligaments and other connective tissues are increasingly loaded. This may explain the increased pain experienced by the participants used by Luethi et al. (1986). The occurrence of injury may then be observed. Although, the increase in pain may result from increased inversion to attenuate impact forces, risk factors associated with lateral ankle injuries in soccer are commonly associated with changes in the surface condition and/or movement speed. This can cause a sudden increase in the horizontal force produced. This can lead to a rapid inversion of the rearfoot, which cause a sudden and excessive load to be applied and an injury to result.

It has been proposed that co-activation of antagonistic muscle groups is an important factor influencing dynamic joint stability to protect the lateral ankle ligaments from the experience of the rapid loading (Baratta, Solomonov, Zhou Letson, Chuinard & D'Ambrosia, 1988). However, it could be speculated that when injury occurs to the lateral ankle ligaments, the magnitude and timing of the force produced by the peroneal muscles may be insufficient to resist excessive movements.

During walking, Hopkins, McLoda & McCaw (2007) observed that with a sudden inversion movements, the reaction time of the peroneal muscles and the time to reach maximum inversion was such that activation of these muscles were able to control excessive movements. This, it could be speculated, is because during the gait cycle there are many muscles that are activated prior to and during the early stance phase in order to stabilise the ankle and foot (Nakazawa et al., 2004). During this phase of gait, the muscle spindle sensitivity increases which may result in increased joint stiffness. Nakazawa et al. (2004) suggested that this is because the muscles experience an increased activity of the gamma motorneuron and a decreased reaction time as the more sensitive muscle spindle system would trigger a reflexive response with less of a stimulus. Thus, when the muscle spindle sensitivity is enhanced, the receptor would fire at a lower threshold. This would result in an overall decrease in the time from stimulus (sudden, unexpected inversion) to muscle activation (Hopkins et al., 2007) and protect the ankle when the stimulus is unanticipated. However, Nakazawa et al. (2004) suggested that it is the rate of force development that is a critical factor, and that the electromechanical delay of the peroneal muscles reaction would indicate whether the ankle was sufficiently stiff to protect the lateral ankle ligaments. Hopkins et al. (2007) suggested that the peroneal muscles may act in a timely fashion to respond to a sudden inversion during walking but questioned whether the stiffness provided by the contraction would be sufficient enough to provide stability to the joint. They also suggested that their findings were only relevant to walking and that during running and turning the rate of inversion would likely be greater. In such situations, the peroneal muscles activation may not be rapid or strong enough to control the movement, contributing to greater risk of injury. However, Hopkins et al. (2007) cited the comments of Johansson et al. (1991) who stated that although it has been widely accepted that injury results when the muscle response following the initiation of movement is too slow, the changeable state of muscle stiffness at the time of ankle displacement may also be influential. Hopkins et al. (2007) therefore suggested that, since both the muscle pre-activation and resultant muscle stiffness will change according to the sensory information from the joint and muscle during movement, ignoring the state of the muscle at the time of joint movement may provide a too simplistic explanation of the contribution of peroneal muscles to injury occurrence. As such the muscle pre-activation and muscle stiffness may be reduced when lateral ankle ligament damage occurs.

Once these ligaments are subjected to such a magnitude or rate of loading that they become damaged, the stability about the joint is reduced. In these cases the natural support provided by the ankle ligaments is lost and the ankle must remain stiff through greater muscular control of the peroneal muscles to enable the ligaments to heal. This prevents excessive inversion but also reduces the natural attenuation provided by moderate levels of inversion (Dayakidis & Boudolos, 2006). As such, the VGRF at impact in 'stable' and 'unstable' ankles in the same athlete has been shown to be significantly greater in the more unstable ankle, with a more rapid onset during the first millisecond post impact, whilst the medial aspect of force remained unchanged (Dayakidis & Boudolos, 2006). Therefore, increased muscular stability controls the horizontal forces to maintain joint stability, but may predispose the athlete to injury through a high VGRF at impact. The same could be said if the level of inversion is restricted in the first instance by using a device such as an ankle support. By placing a large amount of support around the ankle, the movement of the joint may be limited. In this case, the external force is increased due to the shortening of movement range which is needed to attenuate peak VGRF at impact and loading rate. The external force must therefore be mostly absorbed by the bones and joints, which create greater compressive forces in the joint and can correspond to greater pain (Luethi et al., 1986). It has been proposed that human joints are adapted for certain ranges of motion, where too much or too little movement at the joint can disturb the equilibrium of force absorption by the various elements to bring about ankle injury (Luethi et al., 1986).

Along with the subtalar joint, the ankle-foot complex includes the talocural joint. The movement of this joint also acts as an important shock attenuation or dampening mechanism (Gross & Nelson, 1988). The ankle is a synovial hinge joint, and as described in section 2.3.3, the tibia and fibula create a deep socket or mortise, that contains the talus (Whiting & Zernicke, 1998) (see Figure 2.2) and many associated ligaments (Norkin & Levangie, 1992). This joint is generally considered to have a single oblique axis with 1° of freedom, where the foot is free to move in directions described as dorsi-flexion and plantar flexion, which occurs mostly in the sagittal plane. In a dorsi-flexed position, the talus fits snugly within the mortise and is quite stable. As the ankle plantar flexes, the narrower posterior section of the talus rotates into the area between the malleoli. This causes a looser joint fit, which compromises joint stability (Whiting & Zernicke, 1998), and allows the subtalar joint to enter greater ranges of movement, placing greater strain on the ankle ligaments.

When the ankle is in a neutral position the foot is at a right angle to the tibia. The ankle joint axis passes approximately through the fibular maleolus and the body of the talus, and through or just below the tibial maleolus. The fibular maleolus however, extends more distally than the tibial maleolus and is positioned in a more posterior direction. This more posterior position of the fibular maleolus is due to the normal torsion or twist that exists in the distal tibia relative to its proximal plateau. The distal tibia is twisted laterally compared with its proximal portion, producing a toe-out position of the foot in normal standing. Given the position of the two malleoli, the axis of the ankle is considered to be rotated laterally 20 to 30° in the transverse plane and inclined 10° down on the lateral side. This inclination of the ankle joint axis results in motion across two planes: dorsi-flexion of the foot brings the foot up and slightly lateral (increased toe-out), whereas plantar flexion brings the foot down and medial (decreased toe-out) (Norkin & Levangie, 1992) (Figure 2.12).

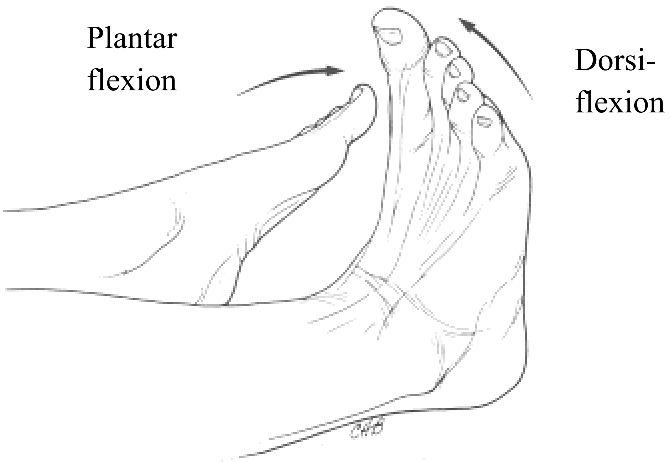


Figure 2.12. Representation of foot-ankle joint complex dorsi-flexion and plantar flexion (Adapted from Miller, 1990 with permission of the author)

During running, the normal range of motion of the talocural joint is generally 20° of dorsi-flexion from neutral, and 30 to 50° of plantar flexion from neutral. The measurement of plantar and dorsi-flexion of the ankle joint has been widely presented in the literature. Maximal dorsi-flexion occurs during the stance phase of gait (Johanson, Baer, Hovermale & Phouthavong, 2008). However, the term ankle dorsi-flexion implies the measure of the talocural joint in isolation. Instead, ankle dorsi-flexion comprises of talocural, subtalar and mid-foot joint movement. Thus, restriction at any of these joints may limit total ankle movement (Lundberg, Goldie, Kalin & Selvik, 1989). Lundberg, Goldie, Kalin & Selvik (1989) used cadaver data to study the contribution of joints between the tibia and the first metatarsal on ankle joint movement, including the subtalar joint, and the talocalcaneal joint. This was performed under full body load and was analyzed using the controlled and systematic 10° change of motion of the foot from 30° of plantar flexion to 30° of dorsi-flexion. It was found that for 10° of total ankle movement, the talocrural joint accounted for only 5.9° (S.D. 2.3°). Therefore, ankle dorsi-flexion, is not exclusively ankle motion (ie, talocrural motion). Instead, subtalar motion is particularly influential, as it also has components of dorsi-flexion in its motion (Scharfbillig & Scutter, 2004). Therefore, this movement could be more accurately described as dorsi-flexion of the foot on the leg rather than simply terming the movement ankle dorsi-flexion (Scharfbillig & Scutter, 2004). This would also suggest that plantar flexion movements would experience similar contributions from these joints.

As previously discussed, plantar flexion is a mechanism which can help control the experience of impact forces. When plantar flexion of the ankle is restricted, the foot is in a less plantar flexed position at ground contact, and larger peak forces which are generated at a much greater rate after ground contact (Brizuela et al., 1997), thereby increasing the loading on the lower limbs. During running tasks, a plantar flexed position may be beneficial to reduce the forces. During lateral movements, inversion occurs and is accentuated by plantar flexion and this inverted and plantar flexed position makes the ankle most susceptible to injury. Therefore, optimal levels are required to maximise impact attenuation whilst maintaining the stability of the ankle-foot complex.

The magnitude of dorsi-flexion may also influence the risk of injury. Runners with limited ankle joint dorsi-flexion are regarded as having a predisposition to lower limb injuries including, muscle strains (Ekstrand & Gillquist, 1982), plantar fasciitis (Riddle, Pulisic, Pidcoe, & Johnson, 2003), Achilles tendinopathy (Kaufman, Brodine, Shaffer, Johnson, & Cullison, 1999; Wilder & Sethi, 2004), stress fractures (Neely, 1998; Wilder & Sethi, 2004), shin splints (Neely, 1998; Messier & Pittala, 1988; Wilder & Sethi, 2004), iliotibial band friction syndrome (Neely, 1998; Messier & Pittala, 1988), and patellofemoral syndrome (Lun, Meeuwisse, Stergiou & Stefanyshyn, 2004). Likewise, in “normal” runners, reduced levels of dorsi-flexion have been associated with a lower risk of Achilles tendon injury (Dixon & Kerwin, 1999).

Kinematic calculations and measurement techniques

To assess sagittal plane kinematics, such as ankle joint movements, markers have been used to signify body land marks or centres of rotation, although these are only approximations (Bartlett, 1997; Dixon, 1996; Milliron & Cavanagh, 1990). Using these markers, however, allows segments to be defined and relative angle between them to be identified. The ankle joint angle is measured to determine the magnitude of ankle plantar flexion and dorsi-flexion (movement of the foot relative to the lower leg, not just the talocural joint). To calculate this movement, the most proximal marker is positioned at the knee. A marker is then placed at the ankle, and the shank segment is defined as the line joining these markers. However, Milliron and Cavanagh (1990) showed that the knee joint centre is difficult to establish as it changes throughout the gait cycle. They suggested that the use of one marker will typically result in the incorrect placement position and that the location of the centre of the knee rotation is misguided most of the time. The ankle marker is typically placed on the lateral maleolus to represent the ankle joint centre. Similar to the knee, the ankle joint centre also moves throughout the gait cycle but does not impact on the results significantly (Engsberg, 1987; Siegler, Chen & Scheck, 1988). However, in comparison to the knee, the bony landmarks and the small amount of skin movement at the ankle makes marker placement easier (Milliron & Cavanagh, 1990). To define the second segment, the line of the foot is used. The definition of the foot segment differs between research publications. Either the foot segment is defined as the segment from the ankle joint marker to a marker placed on the fifth metatarsal joint (Dixon & Stiles, 2002; Dixon & Kerwin, 1999), or the segment between a marker placed on the heel and the marker defining the MTP joint (Stiles & Dixon, 2006; Nigg & Bahlsen, 1988; Zhang, Clowers, Kohstall & Yu, 2005). Like the ankle, the location of the fifth metatarsal is more easily defined than the knee even when a shoe is worn (Milliron & Cavanagh, 1990). The ankle angle is then calculated as the relative change between the segments.

The kinematics of the knee can influence the forces during the gait cycle. This will be discussed in section 2.5. To define the angle of the knee, it is common for participants to have a marker placed on the hip, normally on the superior border of the greater trochanter, although identifying the hip joint centre can be difficult in some individuals due to high subcutaneous fat deposits and skin movement in the region of interest

(Milliron & Cavanagh 1990). As well as defining the shank, the knee joint centre of rotation is also used to define the distal end of the thigh segment. The thigh segment is then measured relative to the shank segment to calculate the knee angle.

The quantification of the pronation/supination of the subtalar joint is particularly important due to the role it can play in increasing one's susceptibility to an overuse or acute injury. As described previously, eversion movements are commonly used as an accurate estimate of pronation (Clarke, Frederick & Hamill, 1984). Likewise, the use of inversion is an appropriate measurement of supination. To establish an approximation of the inversion and eversion motion, the rearfoot angle is calculated from the varus or valgus of the calcaneous with respect to the tibia (Clarke, Frederick & Hamill, 1984; Whiting & Zernicke, 1998), whilst in the frontal plane and from behind. The component of calcaneal eversion is seen as an increase in the medial angulations between the long axis of the tibia and the axis through the tuberosity of the calcaneus. The most common method, in which rearfoot motion is analysed, is by the use of four markers. Two markers are placed on the proximal and distal midline of the Achilles tendon during standing. Another two markers are placed on the proximal and distal calcaneus to define the rearfoot (Clarke, Frederick & Hamill, 1983, 1984; Soutas-Little, Beavis, Verstraete & Markus, 1988) (Figure 2.13).

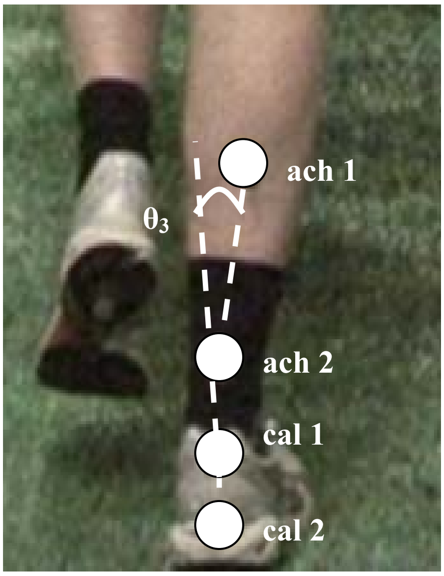


Figure 2.13. A posterior view of the marker positioned at the proximal and distal calcaneous (cal 1 and cal 2) and two defining the line of the tibia bisecting the leg in the frontal plane (ach 1 and ach 2). The angle between these segments is used to assess rearfoot movement.

When calculating inversion and eversion angles, care must be taken when placing the markers. Cavanagh (1990) demonstrated that small variations in the medial/lateral placement of a marker may significantly alter the calculated angles. A difference in marker placement of 3 mm can change the rearfoot angle by more than 4°. Even greater differences were shown when shoe markers were placed close together where the rearfoot angle increased to at least 8°. Therefore, the repeatability of the marker placement is essential for comparisons to be made across studies. Therefore, the correct marker placement is paramount for calculation of accurate and reliable marker co-ordinates (Dixon, 1996). Likewise, a 1 cm difference in marker position of the hip and knee marker can have up to a 16° influence on the calculated knee angle (Milliron & Cavanagh, 1990). This variability can have a large influence on the differences of angles found between subjects. This can result from inaccurate marker placement or differences in the landmark locations. Variability in marker placement can also influence the angles calculated on the same person on repeated occasions. However, despite the importance of accurate and consistent marker placement, in practice this can be difficult to standardise. To reduce the effect of these variations, standing values can be collected so that the kinematic data can be referenced to this neutral position and any disparity in marker placement between conditions has little impact on the outcome data.

The movement of the calcaneus within the shoe during running has also been highlighted as a concern for the accuracy of angle measurements (Clarke, Lafortune, Williams, & Cavanagh, 1980; Nigg, Bahlsen, Denoth, Luethi, & Stacoff, 1986; Stacoff, Reinschmidt & Stüssi, 1992). However, the movement of the calcaneous and shoe markers are similar, although not identical. The movement that did occur between the shoe and rearfoot was systematic, indicating that the movement of the rear shoe and posterior aspect of the lower leg can be used to represent the relative movement of the calcaneus and the lower leg (Dixon, 1996). However, Clarke, Frederick and Hamill (1984) highlighted concerns regarding the additional movement error can occur if the shoe does not fit sufficiently. These finding were shown when running in running shoes. As soccer boots are designed with less cushioning material around the mid-sole and upper than running shoes, a tighter fit is provided by the soccer boot. This offers greater support compared to running trainers, and therefore the movement of the rearfoot relative to the shoe would be expected to be similar if not less in a soccer boot.

Finally, interpretation of kinematic data has to be done with care as humans are complex biological organisms with inter-individual anatomical and functional differences (Bates et al., 1983; Luethi et al., 1996). Consequently, there is large variability in kinematic measurement between subjects and this could effect the observation of significant differences. Due to the large variability in running kinematics between subjects, it may be appropriate to present data for individual subjects, eliminating the possibility of obscuring the behaviour of individuals.

Historically, kinematic variations occurring during locomotion have been measured by collecting the coordinate data of key points of interest using high speed cinematography, which allows frame by frame assessment of a specific action (Cavanagh, 1987). The use of video is more advantageous than cine-film, which was used previously to collect kinematic data, because of its low cost, easy use, and the immediate availability of the recordings. To obtain co-ordinate data, researchers perform manual digitisation of frames on cine-film and video, which means that for each frame, the key points of interest are identified and marked by hand. From these points, body segments can be determined by joining the appropriated points together and angles can be calculated between the segments. However, the accuracy of the digitisation is related to the knowledge and experience of the person performing the digitisation. Also, the quality of the digitisation is influenced by the quality of image resolution, and the fatigue and boredom that is evident with repeated digitisation. The use of automatic tracking technology has been employed more recently for the collection of kinematic data, which avoids the use of manual digitisation and removes these inherent problems. This has been shown to be an accurate and reliable method (Stiles, 2005). Commonly this method uses computer software to track reflective markers worn by the participant. However, this technology is more expensive than the use of traditional video cameras and often requires a more complex set up that is not always practical for some sporting situations.

The data collection frequency is a very important consideration to be made when performing kinematic analysis. A variety of different frequencies are often used to collect data in sports and exercise biomechanics. To choose the correct collection frequency, the sampling theorem is commonly used, which suggests that the process signal must be sampled at a frequency that is at least twice as high as the highest frequency present in the signal itself (Robertson, Caldwell, Hamill & Whittlesey, 2004).

McGinnis (2005) explained that this minimal sampling is known as the Nyquist sampling frequency. In human motion the highest voluntary frequency is less than 10 Hz, so a 20 Hz sampling frequency should be satisfactory. In reality, biomechanical research uses sampling frequencies of between 5 and 10 times the highest frequency in the signal (McGinnis, 2005). However, McGinnis (2005) highlighted that for movements that require impacts or rapidly changing states, a much faster sampling frequency is required. This ensures that the signal is accurately portrayed without missing peak values.

Kinematic data can be obtained using one, two or more cameras. The number of cameras depends on the type of movement being performed, the accuracy desired by the researcher, and the plane of movement in which the movement takes place. These planes can be referred to as the frontal, sagittal and transverse planes (Bartlett, 1997) (Figure 2.14).

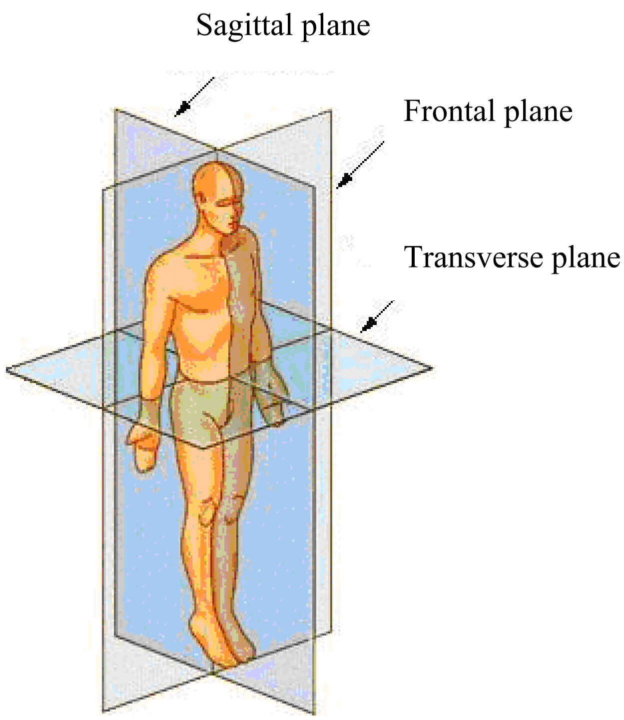


Figure 2.14. Diagram identifying the sagittal, frontal and transverse planes of motion (Adapted from Birmingham, 2009)

To calculate the movement patterns occurring in these planes, body segments or bony landmarks are defined, often with the use of markers, and scaled coordinate of these markers are produced. However, it is apparent that two-dimensional analysis using one camera is a limiting factor when studying complex human motion (Shapiro, 1978). For example, when using 1 camera, perspective error can be introduced into two dimensional data. This occurs when the image is scaled using an object that is parallel to the photographic plane but where the scaling object is situated away (i.e. further forward or back) from the movement plane (the plane of performance) where the participant in positioned (Figure 2.15). This influences the size of the projected two dimensional image of the participant relative to the scaling object, causing an apparent discrepancy in length between two objects that are equal in length. An example of this is when two limbs are measured but where one of the limbs is closer to the camera than the other. The term is also used to refer to error in the recorded length of the limb or between segments that are at an angle to the photographic plane and therefore appear shorter than they actually are. This can occur when measuring the outward position of the toe at contact. In both these cases the object is perpendicular with the line of the optical axis, i.e. directly in front of the camera. On the other hand, parallex error is introduced when viewing the markers away from the optical axis, such that the plane of motion is not perpendicular with this optical axis (a) and (b) (Figure 2.15).

The three-dimensionality of a marker location can also induce error and requires the two-dimensional analysis of the marker to be performed with care. Error can be introduced if there is any non-alignment of the movement plane and the plane perpendicular to the optical axis of the camera (photographic plane). This can often occur if calibration is performed with a simple scaling object in the plane of motion (Bartlett, 1997) (Figure 2.16). Finally, errors in the digitised co-ordinates can be caused through lens distortion, other internal optical camera factors, and the representation of joint axis of rotation based on the estimates of superficial skin markers as well as camera vibration, image blurring and low resolution (Bartlett, 1997).

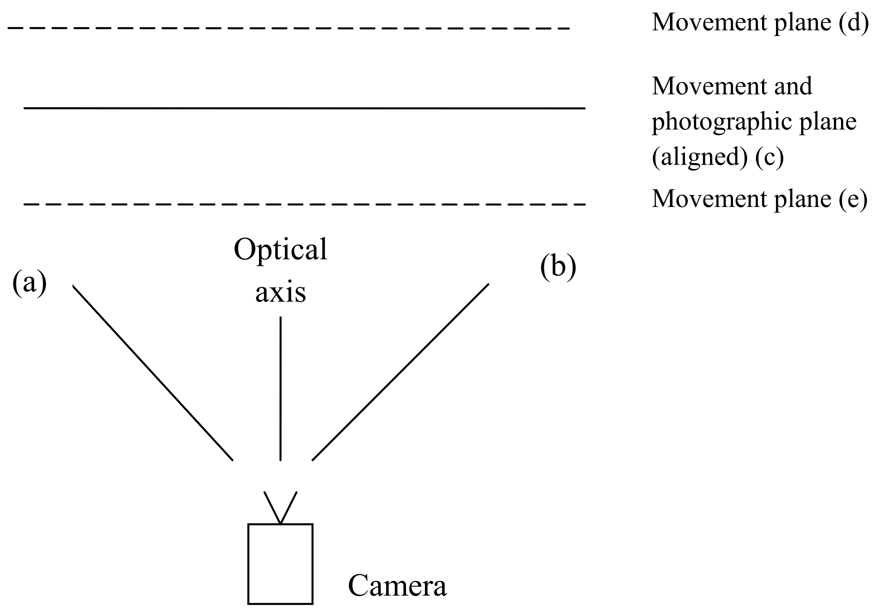


Figure 2.15. Example of camera position, movement plane, photographic plane and optical axis when perspective error (running in movement plane d and e, whilst the photographic plane is c) and paralex error (angles taken at positions a and b whilst the optical axis is perpendicular to the movement plane) is introduced.

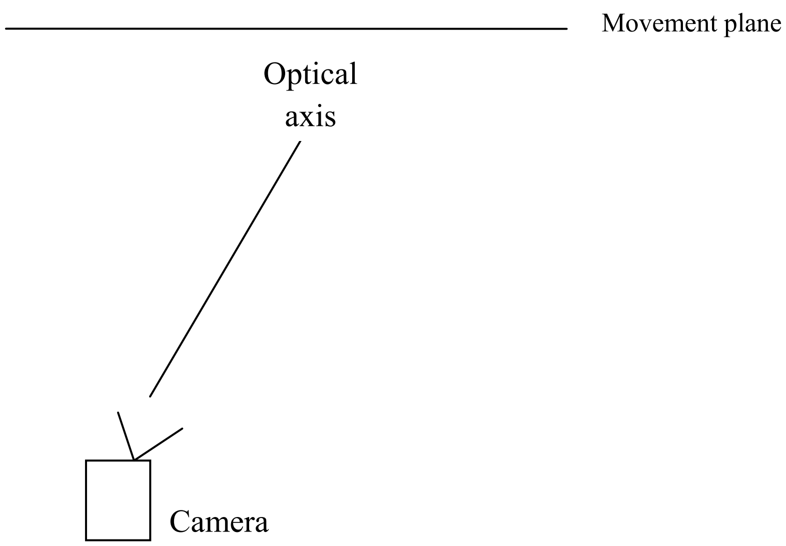


Figure 2.16. Example of camera position for the non-alignment of optical axis and movement plane.

Many of the methodological issues that arise with two dimensional analyses can be removed if three dimensional analyses are performed. With this form of analysis, the angles between body segments can be calculated accurately in the three global reference planes, whilst also allowing the calculation of the other angles that can not be observed with a single camera (Bartlett, 1997). Several algorithms are available to reconstruct the three-dimensional movement space coordinates from two or more images that take into consideration changes in movement perspective. However, most require implicit reconstruction of the line (ray) from each camera that is directed at a point of interest, such as a skin marker. The location is then estimated to be that which is closest to the two rays. However, these techniques are commonly too restrictive for use in sports situations (Bartlett, 2007). Shapiro (1978) described that there are several other methods that have been developed using stereo photogrammetric techniques to reconstruct images into three-dimensions and have yielded excellent results in terms of their accuracy. However, a major problem with these multi-camera methods is the need to measure all the factors related to the external camera parameters. Therefore, they require special metric cameras with known parameters to be used (Shapiro, 1978).

Abdel-Aziz & Karara (1971) developed a methodology that would allow the application of the stereo-photogrammetric technique to situations in which non-metric cameras (i.e. those in which the internal orientations are unknown) are used (Shapiro, 1978). The direct linear transformation (DLT) method uses two or more cameras to provide two sets of two dimensional coordinates of an image, that combine to calculate a set of three dimensional coordinates. This method of reconstruction is commonly used in the biomechanical literature, due to the accuracy of the results with non-metric and still (photography) cameras (Marzan & Karara, 1975 cited in Shapiro, 1978) as well as with high speed cinematography (Shapiro, 1978). During calculation of the joint co-ordinates with two cameras, each camera is required to see all the joint markers of interest at all times. Therefore, the cameras are set up so that the experimental field is covered by the over lap of camera angles (Shapiro, 1978). As such, the technique is also preferred to other techniques due to the great flexibility in camera set up (Shapiro, 1978; Woods & Marshall, 1986). During some movements it is not always possible to view all markers at all times with the use of just two cameras. Therefore, another advantage of the DLT method is that more cameras can be added. However, there is one prominent limitation of the DLT method. Once the calibration object coordinates have been used to calculate the DLT coefficients, the position of the camera must not change unless more complex corrections are to be used. Establishing appropriate positions for the cameras so that all markers can be seen is therefore an important consideration that must be made prior to data collection.

To transform the image coordinates to a movement space coordinates system, camera calibration involving 11 or more transformation parameters/coefficients (L1-L11) is required for each camera independently (Bartlett, 2007). The DLT uses the measured two-dimensional image coordinates (x, y) of a calibration object from each camera, along with the known coordinates, to estimate these DLT transformation parameters. Commonly, at least 6 calibration points which are each visible from two or more cameras are used to provide these parameters (Bartlett, 2007). Each marker has three dimensional coordinates, (X, Y, Z), known with respect to three mutually perpendicular axes. This is provided by the use of a calibration object which, when viewed as a two-dimensional image, each point can be clearly and unambiguously distinguished from one another (Bartlett, 2007; Shapiro, 1978). By using these transformation coefficients/parameters and the image co-ordinates, it is then possible to calculate the movement space co-ordinates of an arbitrary point such as a marker worn by a participant which is visible in the calibrated space (Bartlett, 2007).

The DLT equation is presented below. In the equation, *x* and *y* are digitised co-ordinates of a point, δx and δy are the associated DLT errors, X, Y, and Z are the object space coordinates (calibration frame) and L1, L2,……,L11 are the unknown DLT parameters of each camera. Up to 11 additional DLT parameters can be included if necessary, to allow for symmetry, lens distortion, asymmetrical lens distortion and non-linear lens distortion (4 parameters) (Bartlett, 2007).

$$x+\delta x = \frac{\mathrm{L1x+L2y+L3z+L4}}{\mathrm{L5x+L6y+L11z+1}} \qquad \text{Equation 2.1}$$

$$y+\delta y = \frac{\mathrm{L5x+L6y+L7z+L8}}{\mathrm{L9x+L10y+L11z+1}} \qquad \text{Equation 2.2}$$

The accuracy of the marker reconstruction is fundamentally important as it influences the accuracy of the angles measured. Camera configuration is one important consideration that must be made with respect to marker reconstruction accuracy. Optimal camera configuration can be calculated on the basis of prior error tolerance

(Shapiro, 1978). It has been indicated that if the distance between cameras, is approximately one third of the distance from the object to the line between cameras and convergence is kept at a minimum, acceptable data can be obtained (Shapiro, 1978). However, Woods and Marshall (1986) highlighted the research conducted by Putnam (1979, cited in Woods & Marshall, 1986) and Neal (1983 cited in Woods & Marshall, 1986), which showed that camera positioning and orientation is not a critical factor for accurate data. For a converging set up with two cameras, the optimal angle of intersection is theoretically 90° (Bartlett, 2007). However, as the camera should be positioned to get the best view of the performer, deviations from this can be tolerated if kept within 60-120º (Bartlett, 2007). In practice, it is customary to express this angle in terms of the distance:base ratio (the ratio of perpendicular distance from the mid-camera point to the participant and the distance between cameras) (Woods & Marshall, 1986). It was shown that the choice of a 1:3 to 2:1 ratio appears to produce equally reliable data (Putnam 1979 cited in Woods & Marshall, 1986; Neal 1983 cited in Woods & Marshall, 1986), although Woods and Marshall did find that on the basis of the reconstruction errors, a camera setup with a base ratio of approximately 1:2 will produce a more accurate result than with a 1:1 ratio.

Chen, Armstrong & Raftopoulos, (1994), Putnam (1979 cited in Woods & Marshall, 1986) and Shapiro (1978) calculated the average co-ordinate reconstruction error with the DLT reconstruction method. These authors each calculated a root mean square error (RMS) to assess the coordinate reconstruction error, and cited that the error was in the range of 1-6 mm for each of the three dimensional global planes. This was stated as comparable to error associated with two dimensional analyses (Shapiro, 1978). However, when manually digitising, the error is not only influenced by the DLT reconstruction, but also the ability of the experimenter to digitise the points, the stability of the camera and maintenance of their fixed points (Shapiro 1978). This error may however, be reduced if an automatic tracking system is used. The accuracy of reconstruction has also been related to the number and distribution of the control points on a calibration frame. When the number of control points is thoroughly and evenly distributed throughout the object space, the accuracy of the reconstruction is thought to be improved (Chen et al., 1994; Frazer, 1982, cited in Woods & Marshall, 1986). The accuracy of the reconstruction of an arbitrary point is also influenced by the accuracy of the known calibration points coordinates (Bartlett, 2007), and become stronger and more reliable when a greater number of well distributed calibration points are used

(Chen et al., 1994; Karara, 1980; Putnam, 1979 cited in Woods & Marshall, 1986;). The addition of a 12th parameter to account for systematic error arising from symmetrical lens distortion is also thought to improve the accuracy further (Putnam, 1979 cited in Woods & Marshall, 1986).

Woods and Marshall (1986) measured the effect of the different factors that influence data reconstruction error. They observed that the average reconstruction error with 30 control points and the use of 11 DLT coefficients and a left-right camera position that equates to a ratio of 1:2 provided a reconstruction was more accurate than a reconstruction using a camera set up with a ratio of 1:1 and when the 12th DLT coefficient was used. The error with a camera ratio of 1:2, and with an 11 parameter DLT was 5.7 mm, which is similar to those reported by Chen et al., (1994), Shapiro, (1978) and Putnam (1979 cited in Woods & Marshall, 1986). However, it was also observed that the transformation based on the seven control point configurations did not produce significantly different results from configurations of 11 or 30 control points. This contradicts some of the previously stated recommendations (Chen et al., 1994; Karara, 1980; Putnam, 1979 cited in Woods & Marshall, 1986). It has also been concluded that accurate reconstruction can only be guaranteed within the calibration volume, so the distribution of the markers for the calibration volume must be around or within the space in which the sports movement takes place (Bartlett, 2007). Shapiro (1978) reported little loss in measurement accuracy at the extremes of a photographic field which were not covered by control points. However, Woods and Marshall (1986) pointed out that the errors were 2-4% (in a 1m field), which would be considered by some as unacceptable. Woods and Marshall, (1986) then found that in contrary to Shapiro (1978), the errors associated with three-dimensional reconstruction were greatly influenced by the distribution of the calibration points, particularly when the movement markers on the participant lay outside the distribution of the control points. Therefore, Woods and Marshall (1986) suggested that if a compromise must be made, fewer control points that are well distributed throughout the object space will result in better results than extrapolation of the control points to cover areas not covered by the calibration object.

Kinematic data collected by using three-dimensional analysis can remove many inherent problems with marker reconstruction, yet it has been highlighted that the provision of the three-dimensional coordinates of a marker can still only determine two-dimensional angles in the three global reference planes (Soutas-Little et al. 1987). However, many movements occur three-dimensionally, that is they do not occur in just one fixed reference plane but can occur across two or three. This is particularly apparent for the measurement of rearfoot movement. As previously indicated, the change in rearfoot angle is determined by a line joining two marked points representing the proximal and distal Achilles tendon, and measured relative to a line defined by two markers placed on the inferior and superior calcaneous or shoe (Clarke, Frederick & Hamill, 1983; 1984; Soutas-Little et al., 1987). These lines are recorded in the vertical plane. However, due to the angle of the foot during ground contact, particularly during impact and propulsion, these lines do not represent the true length or orientation of the actual lines (Soutas-Little et al. 1987). Therefore, the angles between the lines are not truly reflective of the angles experienced by the participant and are not accurate measures of eversion and inversion except when the subject has both foot and shank in the vertical/frontal plane (Soutas-Little et al., 1987). In addition, because of the alignment of the ankle angle, there is a dependant amount of dorsi-flexion and adduction movement with eversion, which causes a toe out position and moves the projected lines of the rearfoot away from the frontal plane, which further increases the error introduced.

Many attempts have been made to establish a solution regarding representing angles in a three-dimensional space. These techniques have involved establishing a fixed laboratory co-ordinate system and measuring the rotations about these axes. However, these tests yield angles which are sequence dependant, meaning that there is not a unique set of angles to describe the orientation of the body, as the values would change with the order of rotations (Soutas-Little et al., 1987). Soutas-Little et al. (1987) developed a technique where markers were placed on the foot and shank. These included markers positioned as described previously for measurements of two-dimensional rearfoot angle, although the distance was less than the 20 cm recommended by Clarke, Frederick & Hamill, (1983; 1984). These markers were then used to model the foot-shank as two rigid bodies, enabling the joint co-ordinate system to be used in conjunction with three-dimensional target data to determine foot motion. This allows the analysis of motion of the foot about three axes, giving rotations which approximate the motions generally referred to as plantar flexion and dorsi-flexion, inversion and eversion and medial and lateral rotation.

Soutas-Little et al. (1988) compared those angles calculated with the local co-ordinate system to two-dimensional angles, and found differences in angular data for the two anatomical methods. It was reported that the maximum difference in rearfoot eversion was 20% when the projection of two lines, i.e. two-dimensional angles, was compared to the actual angle. However, Soutas-Little et al. (1988) highlighted that these differences depend on the running style and the amount of rotation that occurs between the two axes. Soutas-Little et al. (1988) also reported that the variation in angular velocity and acceleration were greater than expected. This was most pronounced in the angular acceleration of the projected inversion-eversion angle. Soutas-Little et al. (1988) suggested that these differences would have a large effect if the data were used in inverse dynamics calculations. Finally, the evidence presented showed that the difference in the calculation of dorsi-flexion and plantar flexion and medial and lateral rotation angle measurements was negligible between techniques.

In the investigation of the biomechanical risk factors associated with soccer injury to the lower extremity, the ankle-foot complex and knee are important structures associated with different aspects of injury occurrence. The joints of these structures can be measured using joint markers to represent limb segments. Three dimensional co-ordinate data can be collected using kinematic analysis via a DLT which removes some of the inherent limitations associated with kinematic data collection. Three dimensional joint angles can also be used to assess angle that occur across multiple planes, which improves the accuracy of the data obtained. Kinematic data collection contributes substantially to the understanding of injury risk factors, and thus should be included in any biomechanical analysis.

2.3.5.3. Joint moments.

Ground reaction force measurements provide a relatively quick and easy estimate of the forces acting throughout the lower extremity. Vertical ground reaction forces are representative of the passive forces that propagate the lower extremity. High magnitudes of these forces are proposed to result in high force transients that generate stress and strain to the lower extremity joints. However, it has been observed that much of the impact force can be reduced through kinematic adaptations (Stacoff et al. 1988; Perry & Lafortune, 1995). In response to kinematic changes, there are greater muscular forces generated through muscular contraction to provide joint control. As such, most force that cross a joint comes from contracture of the muscles that cross the joint (Radin et al., 1973). They are also used during the initiation and maintenance of locomotion. The change in muscular force results from the kinematic change influencing the lever arm length of the force. Along with muscular control, ankle ligaments also contribute to joint stability through passive forces. Further still, Winter (1984) showed that internal muscular load can be sensitive to increased GRF. Therefore, it may be more appropriate to measure these internal forces when trying to assess overuse and acute injuries that occur in soccer. Scott and Winter (1990) found that peak internal forces at mid-stance were more closely associated with a number of injuries, than the measurement of external forces at impact. Likewise, Stefanyshyn et al. (2001) found that despite no significant correlation being observed, the magnitude of internal forces is more closely linked to injury than peak external forces and loading rate, and despite no significant differences being observed, speculated that with a larger sample of participants, significant differences were likely to be observed. Therefore, despite the role of kinematic adaptations to lower excessive force transients and the risk of impact related injuries, the risk of muscular, tendinous and ligamentous loading may be accentuated from the attenuation of these impact forces and contribute to injury. Further still, internal forces are also sensitive to small changes in external forces (Winter, 1984). Therefore, it may be more appropriate to measure joint moments or internal forces present in specific structures, than simply measuring VGRF.

A representative estimation of internal joint loading can be obtained by the calculation of a resultant joint moment (Stefanyshyn, 2003). This represents the net effect of all agonist and antagonist muscle, tendon and ligament activity (Winter 1980; Stefanyshyn, 2003), and the integration of all the neural control acting at each joint (Winter, 1980). The measurement of joint moments is thought to provide the most valuable insight in the assessment of human movements (Winter, 1980). To determine the net moment that occurs about the ankle-foot complex, inverse dynamics (Winter, 1983; Scott & Winter, 1990) and quasi-static methods have been developed (Burdett, 1982; Morlock & Nigg, 1988).

Inverse dynamics is a technique that uses GRF measurements alongside the inertial forces (where the force is proportional to the acceleration of the segment but opposite direction), mass and COM of the segment of interest. The calculation also uses gravitational forces, centre of rotation and the centre of pressure (COP), to calculate the moment found within a joint and estimate the force applied by the muscles to cause or control the movement (Norkin & Levangie, 1992). In the calculation of ankle loading, the quantification of the foot mass, COM and the moment of inertia of the foot segment about the principal axes have been calculated with different methods (Dixon, 1996). Using experimental techniques Dempster (1955), Chandler, Clauser, McConville, Reynolds, & Young (1975) and Clauser, McConville & Young, (1969) calculated the mass, volume, density, and COM values for selected body segments, and for the total body mass using human cadaver data via the pendulum technique. There are other experimental methodologies available to calculate these characteristics, which include volume contour mapping, computerised axial tomography, the quick release, relaxed oscillation and gamma-scanner methods, and magnetic resonance imaging (MRI) (Nigg, 1999). To define each segment, Clauser et al. (1969) and Zatsiorsky et al. (1990 cited in de Leva, 1996) used bony landmarks to identify the segment end points rather than joint centres. De Leva highlighted that some of these landmarks were markedly distant from the joint centers currently used by most researchers as reference points. As such, adjustments have to be made to the original measurements if joint centres are to be used in the calculation (Hinrichs, 1990; de Leva, 1996). Hinrichs (1990) also reported that the new proportions calculated are markedly different than those originally reported, and cautioned against using the original proportions when using joint centres as segment endpoints.

Hatze (1980) wrote that although experimental techniques appear appropriate for determining segment mass, COM and volume, theoretical methods are more suitable for the calculation of the principle moments of inertia and orientation of the principal axis since measurements are taken from individual subjects. In published theoretical methods, the body has been represented as a model and the inertia properties determined mathematically (Hanavan, 1964; Hatze, 1980; Yeadon, 1990). This is generally performed by taking measurements from subjects and imputing their characteristics into the model (Dixon, 1996). Dixon (1996) suggested that the inertia data properties obtained using experimental methods are easy to utilise within inverse dynamics calculations. However, Dixon (1996) also stated that these measurements do not provide data that are specific to the individual participant under investigation, whereas, theoretical methods provide personalised inertia data, but require the time-consuming process of taking measurements from individual subjects.

To choose the appropriate method of determining the inertia characteristics for use within the inverse dynamics calculation, one must consider the requirements of the study (Dixon, 1996). If absolute joint, muscle or tendon forces are to be calculated, accurate inertia data for a subject is important (Nigg, 1999). In contrast, the accuracy of the inertia characteristic measurement is less important if the aim of the study is to compare results of one subject under different conditions (Nigg, 1999). The choice of method will also depend on the subject characteristics. For example, if the subject is an athlete, inertia data measured from young people is likely to be more suitable than cadaver data obtained from relatively old, untrained individuals (Dixon, 1996).

The choice of inertia data has been shown to depend upon the joint on which the moment is calculated. Accurate inertia data is less important for ankle joint moments than for moments about joints higher up the body (Dixon, 1996), as inertia has a negligible effect (less than 2 N.m or 1%) on peak ankle moments during running (Alexander & Vernon 1975). The foot also has a small mass and throughout the stance phase of running, experiences small accelerations (Alexander & Vernon 1975). Some authors have therefore deemed inertia and segment acceleration calculations as unimportant and have excluded them from the calculation of joint moment. These moments are termed quasi-static. Alexander and Vernon (1975) described that the effect of inertia is likely to be large immediately following ground impact, indicating that quasi-static methods may not be suitable for movements calculated during the initial impact phase. However, Morlock and Nigg (1988) found that even during impact with the ground, when accelerations are high, the effect of omitting inertia was negligible on the net moment about the ankle joint in running.

In soccer, the risk of Achilles tendon injury is high during the preseason. Information on the forces produced by individual skeletal muscles or groups of muscles is important for understanding muscle mechanics during normal locomotion (Komi, 1990), and may explain the reasons behind the increased risk of injury. Ankle plantar flexion moment has been used as a measure of the forces occurring by the triceps surae muscle group as they are the main contributor towards plantar flexion (Alexander & Vernon, 1975; Scott & Winter, 1990). Alexander & Vernon (1975) suggested that this muscle group contributes 65 to 100% towards this movement, whereas Scott and Winter (1990) suggested that this contribution is 85%, based on muscle physiological cross sectional area. Since it is often assumed that during ground contact there is a consistent contribution, any difference in values would have a systematic effect on the maximum Achilles tendon force (Dixon, 1996).

The shape of a moment-time history depends upon the movement direction of the segment of interest. A joint moment tending to cause flexion is generally given a positive sign and a moment tending to cause extension a negative sign. An anticlockwise ankle joint moment causing ankle dorsi-flexion is taken to be positive and an anti-clockwise moment causing plantar flexion negative (Dixon, 1996), although the opposite configuration can be used. It has been demonstrated that beyond the first 10% of stance of running gait, there is a negative joint moment about the ankle joint (Winter, 1983; Scott & Winter, 1990), which theoretically can be balanced mathematically by the use of a single muscle group acting to cause plantar flexion (Dixon, 1996). This occurs to control the dorsi-flexion joint movement, and is termed the muscle moment. When running, the peak moment occurs during mid-stance whilst the toes are on the ground (Alexander & Vernon, 1975), and represents the maximum eccentric load being applied to the Achilles tendon. The magnitude of the peak moment is usually about 200 N.m for both two-dimensional (Alexander & Vernon, 1975) and three-dimensional (Reinschmidt & Nigg, 1995) moment calculations. In practice, the main plantar flexion muscle group of the ankle joint is the triceps surae group. However, other muscles have the ability to contribute to the plantar flexion moment, and these include the plantaris, flexor hallucis longus, flexor digitorum longus, tibialis posterior, peroneal muscles (longus and brevis) (Dixon, 1996).

Although the calculation of an ankle moment can be used to indicate ankle loading, it does not accurately describe the forces of the triceps surae muscle group and Achilles tendon. Several investigators have calculated the forces transmitted by the Achilles tendon during walking, running and jumping using in-vivo techniques. Buckle transducers have been surgically implanted under local anaesthesia to record the magnitude of Achilles tendon forces during various tasks (Fukashiro, Komi, Jäirvinen & Miyashita, 1995; Fukashiro, Komi, Jäirvinen & Miyashita, 1993; Gregor Komi Browing & Jarvinen 1991; Komi, 1990; Komi, Fukashiro & Järvinen, 1992). Fushashiro, et al. (1995) collected the in-vivo Achilles tendon forces during jumping. Peak Achilles tendon force was reported as 2233 N, 1895 N and 3786 N during a squat jump, counter movement jump and when hopping respectively. Komi (1990) also measured the in-vivo loading of the Achilles tendon during walking, running and jumping. Amongst the values attained for the different tasks, Komi (1990) highlighted that in some cases peak Achilles tendon force reached as high as 9 kN corresponding to 12.5 body weights (BW) for a single subject performing barefooted running at 6 m.s^{-1}.

Buckle transducers are a valuable tool for measuring in-vivo tendon forces, but have inherent surgical risks, can cause subject discomfort, are expensive, and require long recovery periods, making them impractical for use in most tasks. It is also difficult to attain ethics approval to use this technique. Instead, in-vitro Achilles tendon forces have been estimated during walking and running by using joint moments and the moment arm distance between the joint centre and the line of action of the Achilles tendon. Burdett (1982) calculated the Achilles tendon forces whilst participants ran at 4.5 $m.s^{-1}$ (6 minute $miles^{-1}$), and found that peak values ranged between 5.3 and 10 BW. Likewise, Scott and Winter (1990) used quasi-static moments to obtain maximum Achilles tendon force values that corresponded to a average of 8.2 BW, for three subjects performing running trials at unspecified speeds. Giddings, Beaupré, Whalen & Carter (2000) also calculated Achilles tendon force using quasi-static moments and a finite element model of the foot. Peak forces were reported as 3.9 and 7.7 BW for walking and running respectively.

The use of in-vitro methods has been supported by the research conducted by Fukashiro et al. (1993) who compared directly measured Achilles tendon force with estimated values collected whilst subjects performed jumping activities. The jumps used were maximal vertical jump, with and without counter movements, and a two-legged sub-maximal hopping at a preferred rate. The results showed that the in-vitro technique caused an overestimation of peak forces of 7.0, 24.6 and 3.3% respectively. A similar finding was published by Magnusson, Aagaard, Rosager, Dyhre-Poulsen & Kjaer (2001). Achilles tendon force was estimated and calculated during graded voluntary 10 seconds isometric plantar flexion efforts measured with isokinetic dynomometry. During each contraction, synchronous real-time ultrasonography of aponeurosis displacement, electromyography of the gastrocnemius, soleus and dorsiflexor muscles, and joint angular rotation were obtained. The tendon cross sectional area and moment arm were collected MRI scans. Force and electromyography data from dorsi-flexion efforts were used to examine the effect of coactivation. The tendon force during this task resulted in 3171 N (S.D. 201) corresponding to 2.6% less than the estimated force when co-activation of the other muscles was accounted for. This finding supports the earlier overestimation reported in research published by Fukashiro et al. (1993).

As suggested by Magnusson et al. (2001), some of the inaccuracies may relate to the co-activation of other muscles. Since it is assumed that all force used to control dorsi-flexion results from triceps surae muscle contraction, Achilles tendon calculation would result in an over-estimation of peak forces if some of the net moment is contributed by other muscles. Another key consideration in musculo-skeletal modelling is the accurate determination of the moment arm which is crucial for estimating realistic muscle forces (Maganaris, Baltzpoulos & Sargent, 1998). The calculation of the moment arm from the ankle centre of rotation to the Achilles tendon depends on the accurate determination of the ankle joint centre and the line of action of the Achilles tendon. It is evident that in previous research literature a fixed centre of rotation has been used to indicate the centre of rotation of the ankle joint, which was represented by a marker on the most prominent point of the lateral maleolus in the sagittal plane (Dixon & Kerwin, 1998; 2002; Scott & Winter, 1990). However, it has been demonstrated that the centre of rotation of the ankle joint is not at a fixed point (Engsberg, 1987; Siegler et al., 1988), although this does not significant influence the moment arm lengths obtained (Gregor et al., 1991; Rugg Gregor, Mandelbaum & Chiu, 1990). The use of a fixed ankle joint centre of rotation for measurement of Achilles tendon moment arm length therefore seems appropriate (Dixon, 1996).

In calculating the moment arm between the ankle joint centre and the line of action, several authors have assumed that the moment arm length is constant throughout the range of ankle movement (Baumann, 1981; Morlock & Nigg 1991). However, using anatomical data, a change in Achilles tendon moment arm length has been observed with a change in ankle angle (Burdett, 1982; Dixon & Kerwin 1998; Scott & Winter, 1990). Therefore, since the moment arm about the ankle to the Achilles tendon changes throughout mid-stance, the use of a fixed value introduces an unacceptably large level of error into the calculation. As a result, the accurate determination of a moment arm length over the range of joint angles experienced in running is clearly an important feature in calculating Achilles tendon forces.

The magnitude of error introduced by the use of constant moment arm lengths was investigated by Rugg et al. (1990) who used the MRI to calculate the Achilles tendon moment arm length and the degree of error in moment arm estimations between the techniques. A variety of foot positions and ankle angles were tested, and sagittal plane images of the bones, muscles, tendon and ligaments were obtained. Rugg et al. (1990) found that there was a 20% change in moment arm length when moving from a position of maximum dorsi-flexion to maximum plantar flexion. Assuming there is no change in net ankle moments, a 20% error in moment arm measurement will result in a 20% error in Achilles tendon force calculation. Thus the use of a constant moment arm length will clearly result in significant errors in subsequent force calculations (Dixon, 1996).

To calculate an accurate moment arm on a human participant in biomechanical studies, researchers require an accurate representation of the Achilles tendon line of action. However, there is some debate over the best method to establish this line, as it is dependent upon the insertion and origin of the Achilles tendon. In an attempt to establish the line of Achilles tendon pull, Brueggemann (1985, cited in Dixon, 1996) placed a single marker on the tendon insertion point and the moment arm was determined by measuring the distance from a marker representing the ankle joint centre of rotation to the Achilles tendon marker. This method assumed that the line between the joint centre and Achilles tendon was perpendicular to the line of Achilles tendon action. However, the placement of the single marker did not accurately represent the Achilles tendon line of action and Dixon (1996) suggested that the method introduced large errors into the calculation of the moment arm throughout the stance phase.

Cadaver specimens have been used to obtain information on the point of insertion of the Achilles tendon and the line of pull for different orientations of the foot in relation to the lower leg. Burdett (1982) determined the line of action from the points of origin and insertion of the Achilles tendon from the average cadaver data. The co-ordinates of these points were scaled for each participant using the foot length, medial-lateral malleoli distance and vertical distance from the plantar surface of the foot to the origin of a co-ordinate system. However, a potential problem with the use of cadaver data was demonstrated by Brand, Crowninshield, Wittstock, Pederson, Clark, & van Krieken, (1982), who found that large differences in origin and insertion points existed when comparing small and large cadavers. Likewise, Brand et al. (1982) found that the choice of average or individual cadaver data resulted in large differences in moment arm measurements. Similarly, Burdett (1982) found that the moment arm lengths were significantly different when comparing individual cadaver origin and insertion points to those obtained using average data. Therefore, using this data in order to provide a subject specific line of action is limited to specific participant groups, and the data on muscle points of origin and insertion must be specific to the subject under consideration. Minimisation of error has been achieved either by use of techniques such as x-ray or MRI, by appropriate scaling of measurements, or by use of cadaver data from subjects similar to those under investigation (Dixon, 1996). However, this can be time consuming, expensive and not always possible. Therefore, an alternative method may prove beneficial.

Burdett (1982) showed that throughout the range of ankle movement, the angle between the line of action of the Achilles tendon and the line representing the lower leg was less than 10º (Burdett 1982). Because of this small difference, the line of the participant's lower leg has been used to approximate a parallel line of action of the Achilles tendon (Burdett, 1982; Dixon & Kerwin, 1998; 2002). This line will therefore change orientation and consequently influence the length of the moment arm produced. To evaluate the difference between Achilles tendon forces calculated with a constant moment arm length, and forces calculated when the line of action is approximately the same as the line of the lower leg, Dixon (1996) provided an example whereby Achilles tendon force was calculated using a typical moment arm length of 0.028m, and an ankle to tendon insertion distance of 0.03m, alongside a 10º deviation in the tendon line of action. A 9% difference in estimated Achilles tendon force was observed between the methods. It was concluded that due to relatively small length of the Achilles tendon moment arm about the ankle joint, approximating the line of action as being parallel to the lower leg, which changes throughout stance, can clearly have a marked influence on estimated Achilles tendon forces compared to the use of a consistent moment arm (Dixon, 1996).

Another method of calculating the Achilles tendon line of action was represented by Bobbert, Huijing, & van Ingen Schenau (1986) who used leg length and angular data on the knee and ankle joints to calculate a Achilles tendon moment arm length during a dynamic activity. Using the methods of Grieve et al. (1978) to predict the length of the gastroncemius muscle length from angular data for the knee and ankle joints, Bobbert et al. (1986) established a relationship between the rate of change in the length of the gastrocnemius muscle, the rate of change of the angle between the calcaneus and the lower leg, and the length of the moment arm of the gastrocnemius about the ankle joint. The authors then calculated the moment arm of the gastrocnemius as a percentage of the lower leg segment length. Using this method, the gastrocnemius muscle line of action was assumed to be in a straight line from the point of origin to the point of insertion of the muscle. Since the Achilles tendon is a continuation of the gastrocnemius, the moment arm length of this muscle about the ankle joint was assumed to be equal to the

Achilles tendon moment arm length.

As the calculation of a participant specific line of action seems to benefit the calculation of Achilles tendon force, the studies of Burdett (1982) and Bobbert et al. (1986) provided methods of generating subject-specific Achilles tendon moment arm data which Dixon (1996) used as the premise to investigate different methods of calculating Achilles tendon force. Four models were tested that used a fixed marker at the lateral maleolus and then the measured moment arm in the sagittal plane from this marker to the line of the Achilles tendon action which was calculated in four ways. Model 1 (M1) used the heel as the insertion point of the Achilles tendon and the line of action was defined as acting parallel to the line of the leg, which was consistent with Burdett (1982) but where Burdett estimated the insertion point using cadaver data, Dixon (1996) did not. Instead, Dixon defined this location as the point on the rear most part of the foot when the foot was in a flat position. Model 2 (M2) estimated the line of the Achilles tendon to be between the heel and the knee, and model 3 (M3) used a scaled anatomical model of the foot and lower leg. Finally, as the insertion and origins of the Achilles tendon differ between individuals, model 4 (M4) used points marked on the skin of the Achilles tendon to represent the line of the action, which was estimated as being parallel to the lower leg. Dixon (1996) tested these models for a heel insert and running shoe condition and showed that the use of method 4 yielded the lowest variability in the Achilles tendon force standard deviations. As such, it was suggested that this measurement technique would result in the most reliable data

Table 2.1 *Results of Dixon's (1996) investigation in the reliability of different Achilles tendon force calculations. M1-M4 represents Model 1 to 4. Values are reported as the standard deviations of Achilles tendon force measurements in body weights (BW) between repeated trials*

	Heel Insert	Running Shoe
M1	1.25BW	0.98BW
M2	1.40BW	1.10BW
M3	1.36BW	1.11BW
M4	0.53BW	0.90BW

Dixon and Kerwin (2002) used this model, but also took into consideration the effect that the skin thickness surrounding the Achilles tendon would have on the moment arm. This was calculated for one participant using MRI scans. The results were then used to scale the thickness of the surrounding skin for other the participants. The marker radius was also considered as influential in determining the length of the moment arm and this was removed from the distance before the Achilles tendon force was calculated. The peak forces calculated with this method were comparable to those previously reported in in-vivo studies (Dixon & Kerwin, 2002).

Techniques used to measure Achilles tendon forces are well established. However, very little research has been performed on methodologies which permit the estimation of loads on the lateral ankle ligaments. Peak force tolerance of these structures has been measured on cadaver specimens using in-vivo methods (Butler & Walsh, 2004; Clanton & Porter, 1997; Fujii et al. 2005). However, to the author's knowledge, no in-vitro methods of determining ligament forces have been established. This may be because it is difficult to measure ankle ligament force due to the varying direction of the line of action for each ligament throughout movement. It may also be because the contribution of each ligament to the forces produced during lateral movements varies, particularly when different levels of inversion/eversion and plantar flexion/dorsi-flexion are shown. Instead, the ligaments have been grouped to provide a single combined force, which requires a complex computer model and data to be collected from seated participants (Langer, Komistek, Kane, Dennis, & Mahfouz, 2003). Researchers have also used inverse dynamics to indicate the net moment of all connective tissues that resist rearfoot joint movement during turning and cutting (Park, Stefanyshyn, Lee, & Savage, 2005). By collecting joint moments, inferences are then made regarding a change of load placed on the ankle ligaments and any potential change in injury risk that may occur. Two dimensional eversion moments can be calculated in the frontal plane, although the accuracy of kinematic ankle measurements is significantly reduced in this plane, which influences the accuracy of the subsequent velocity and acceleration derivatives (Soutas-Little et al., 1988). Thus, for accurate three-dimensional eversion moment calculations of soccer players performing lateral movements, three-dimensional movement kinematics are required (Soutas-Little et al., 1987).

The use of internal joint forces measurements via the calculation of joint moments offer an understanding into the internal loading of structures such as the Achilles tendon and ankle ligaments. This loading is sensitive to changes in force and kinematic movements. Thus, where external measurements are unable to detect differences, joint moments may be more revealing. Therefore, when the objective is to evaluate the risk factors associated with injury in soccer, it is important that joint moments are calculated.

2.3.6. Inciting event

The final link in understanding injury causation has been defined as the inciting event, which is the event that takes place leading up to and during the occurrence of injury (Bahr & Holme, 2003). Figure 2.17 shows a simplistic diagram of events leading to injury. The inciting event interacts with the risk factor to bring about unusual loads that cause the injury, and without it injury may not occur.

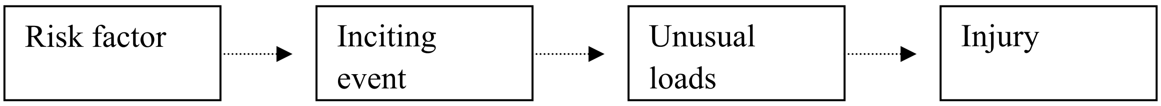


Figure 2.17. A simplistic flow diagram of the sequence of events leading to injury

Soccer is a multi-directional sport, and injuries can occur during the many different movement patterns (Junge & Dvorak, 2004). Traditionally, research into the response of participants to different risk factors has occurred by testing participants during running tasks. However, most non-contact foot injuries occur in soccer whilst twisting and turning as well as running (Wong and Hong, 2005; Woods et al., 2003). Consequently, these movement patterns influence the loading of the lower extremity. These differences change the relative load applied to specific structures, influencing the susceptibility for a structure to become injured. For example, Kaila (2007) found that the magnitude of the internal tibia and valgus moment and anterior joint forces were significantly greater during side cutting compared to a straight ahead running. The movement task may also interact with the risk factor. Where differences between two conditions are not significant during one movement, the variance between conditions has been magnified to cause significant differences during another movement. This has been shown by numerous researchers and thus it may be that during these movements, the risk factors have the largest influence on injury risk (Eils, Streyl, Linnenbecker, Thorwesten, Völker & Rosenbaum 2004; Sims, Hardaker & Queen, 2008; Queen, Hayes, Hardaker & Garrett, 2007; Queen et al. 2008). As such, sports specific movements can be more revealing than running (Stiles & Dixon, 2006). Also, in order to observe differences between risk factors, “highly dynamic manoeuvres” (Coyles, Lake & Patritti, 1998) may be needed in order for differences to be magnified.

Because of the aforementioned considerations, when choosing the tasks in which to test injury risk factors with biomechanical variables, one should consider using a specific task that is being performed when injury results. If this is unclear, a range of movements should be tested as some tasks may not be as revealing of the differences in biomechanical measurements as others. The choice of movement pattern is therefore an important consideration when trying to understand the mechanism behind different risk factors.

2.4. Testing Risk Factors

2.4.1. Playing surfaces.

The effect of playing surface conditions on the aetiology of athletic injury has been discussed in detail in the literature (Arnason et al., 1996; Ekstrand & Gillquist, 1983; Ekstrand & Nigg, 1989; Orchard, 2002; Orchard & Powell, 2003; Woods et al., 2002; Wong & Hong, 2005). Ekstrand and Gillquist (1983) reported that 24% of injuries recognised in soccer were related to the playing surface. Specifically, Ekstrand and Gillquist (1983) and Wong and Hong (2005), identified that inferior quality of the playing surface is one of the most important extrinsic factors influencing the incidence of injury. The quality of the playing surface is dependent upon the environmental conditions, degree of weathering, as well as the management it receives (Ekstrand & Gillquist, 1983).

From the point of view of injuries, low surface deformation/high stiffness properties (cushioning) and both high and low friction properties of a shoe-surface combination, are assumed to be important. These factors are speculated to increase the loads on structures of the musculoskeletal system beyond healthy limits (Nigg, 1990; Nigg & Segesser, 1988; Stefanyshyn, 2003).

2.4.1.1. Surface cushioning.

The number of overuse injuries in soccer is high during the preseason (Woods et al., 2002). This period coincides with the summer months, where the weather has been described as being mainly dry and warm with low humidity (Meyers & Barnhill 2004; Woods et al., 2002). This can cause a drying and hardening of the playing surface. Woods et al. (2002) found that hard and dry playing surfaces were evident during preseason when the rise in overuse injuries occurred. Woods et al. (2002) also identified that for 70% of all Achilles tendon injuries, participation was on a hard surface. This association between hard surfaces and overuse injury is common (Arnason et al., 1996; Ekstrand & Gillquist, 1983; Ekstrand & Nigg, 1989; Orchard, 2002; Woods et al., 2002; Wong & Hong, 2005) and has been speculated to result from increased impact forces (Cavanagh & Lafortune, 1980; Light, MacLellan & Klenerman, 1979). Therefore, the high rate of overuse injuries on hard playing surfaces may be related to an increased VGRF at impact that creates a shock wave that propagates the tibia and transfers from the bone to the soft tissue to cause injury (MacLellan & Vyvyan, 1981). Light & MacLellan (1977) suggested that by reducing the shock wave these distortions can be reduced. Overuse metatarsal injuries in soccer are also common. These have also been linked to hard playing surfaces (Queen et al. 2008), although instead of occurring during impact, these injuries result from increased forefoot loading during the active or propulsive phase of the vertical force-time history. By improving the cushioning, smaller peak forces and pressures may be observable, which may lead to a reduction of injury risk during both impact and propulsion.

To improve the cushioning of natural turf, a quantitative measure of surfaces cushioning is provided via the application of a Clegg hammer (Dr Baden Clegg Pty Ltd, Australia) (Figure 2.18). If correctly performed, the Clegg hammer test can provide a quick and highly reliable method of describing the material hardness of a sample surface and is measured in peak gravities (G) (Nigg, 1990). Testing with a Clegg hammer involves the manual drop of a mass from a predetermined height. Four basic hammer masses are available: 4.5 kg (the "standard Clegg hammer"), 2.25 kg (the "medium Clegg hammer"), 0.5 kg (the "light Clegg hammer") and 20 kg (the "heavy Clegg hammer"). The set drop height for the standard and medium hammers is 45 cm, whilst it is 30 cm for the light and heavy hammers. The 4.5 kg hammer is for "general purpose" used in testing road works, earthworks and airstrips, whereas the two lighter hammers are used primarily for turf or sand testing. The heavy hammer is for testing through a larger zone or on top of the running course of flexible pavements.

Figure 2.18. Image of a 4.5 Kg Clegg hammer

Sifters and Beard (1994) suggested that a natural turf surface can provide a good level of cushioning compared to most other surfaces, as the biomass of natural turf and the associated root zones, provides a uniquely resilient characteristic for cushioning. The level of cushioning provided by natural turf is influenced by the soil texture, moisture content, surface/vegetation cover, and mowing height (Sifers & Beard, 1994; Stiles, Dixon, Guisasola & James, 2008). Dry natural turf with grass cover has received values of approximately 100 G (Sifers and Beard, 1994) and bare turf can score up to 500 G (Dixon et al., 2008; Siters & Beard, 1994). Stiles et al. (2008) found that the peak G differ according to which soil type is being tested, citing Clegg hammer values of between 59.47 G to 72.62 G for natural turf of varying compositions. Stiles et al. (2008) also noticed that with repeated use, compaction of the soil occurred and this increased the surface hardness, but this was dependent upon the soil construction. They observed that ‘rootzone’ representative of modern elite soccer surfaces and ‘heavy’ clay loam surfaces had similar hardness prior to data collection but, following repeated running trials, the rootzone surface possessed a significantly lower level of hardness compared to the clay loam surface.

Due to the association between high VGRF and an increased risk of overuse injury from prospective (Hreljac, 2004; Hreljac et al., 2000) and retrospective studies (Clement & Taunton, 1980; Clement et al., 1981; James, Bates & Osternig, 1978; Munro et al., 1987), as well as the association between hard surfaces with an increased risk of overuse injury (Arnason et al., 1996; Ekstrand & Gillquist, 1983; Ekstrand & Nigg, 1989; Orchard, 2002; Woods et al., 2002; Wong & Hong, 2005), it could be speculated that greater impact forces would be evident during movements performed on a hard surface. This would provide a possible reason for the rise in overuse injuries during preseason. However, although mechanical tests provide a reproducible measure of the surface properties, they do not replicate complex human motion which may change to respond to the surface, and influence the forces experienced. Instead, force plate and kinematic measurements would offer valuable data regarding the cushioning provided to the participant. However, due to the logistical complications of incorporating a natural soil media in the biomechanics laboratory or using biomechanical equipment in the field, progress on the understanding of how humans respond to changes in natural turf properties from the biomechanical perspectives of injury and performance, has been somewhat restricted (Stiles et al., 2008).

Both Dixon et al. (2008) and Stiles et al. (2008) have collected biomechanical and mechanical data in order to understand how the construction of natural surfaces influences the human participant. Stiles et al. (2008) tested the mechanical cushioning of natural turf surfaces that had different soil compositions such as clay, sand and rootzone. They found that the mechanical cushioning was different between each soil composition, although the differences in mechanical hardness did not yield any significant kinematic differences. Instead it was suggested that the maintenance of similar in running geometries were either more preferable when running on these natural surfaces, or that in order to elicit changes in human response during running, the mechanical properties of the natural turf must first be sufficiently different (Stiles et al. 2008).

Dixon et al. (2008) collected plantar foot forces during running on a natural turf away from a biomechanical laboratory by using in-shoe pressure insoles. They found that significant changes in kinetic variables were detected during running on natural surfaces which were of a consistent soil composition and more distinct in their mechanical cushioning, than those reported by Stiles et al. (2008). Dixon et al. (2008) observed significantly lower heel forces on the more cushioned surface. However, several limitations were evident in this research. Firstly, unlike previous research, no kinematic data were collected simultaneously with kinetic data to supplement the research findings. Likewise, a lack of consideration was given to the external validity of the findings, as the density of the surface composition was manufactured within a soil bin, with negligible moisture content, and had no grass cover. However, whilst the ground composition may not have been realistic, the data provided does indicate the potential of soil density to influence kinetic variables when running. Thus, the conclusions provide a point from which our understanding behind the association between natural turf surfaces and injury can be furthered.

2.4.1.2. Surface traction.

In addition to the cushioning of the surface, traction or friction characteristics of the turf can also influence the risk of ankle injuries. As described, the rate of ankle ligament injury is greater during the months of July, August and September (Woods et al., 2003). During this time the surface can be characterised as hard and dry (Woods et al., 2002), which influences the shoe-surface traction. Specifically, the surface-shoe traction will usually have a positive correlation with surface hardness, grass cover and root density (Orchard, 2002; Orchard & Powell 2003). The traction coefficient on turf surfaces also increases with temperature (Torg, Stilwell, & Rogers. 1996) and as the surface ages (Bowers & Martin, 1975), but can be reduced by the presence of moisture (Bowers & Martin, 1975; Orchard, 2002; Orchard & Powell 2003) and other contaminants (Bowers & Martin, 1975). Because of the association between injury and traction, the risk of acute injury has been shown to correlate positively with ground hardness, dryness, grass cover (Orchard, 2002) and temperature (Torq et al., 1996; Orchard, 2002; Orchard & Powell, 2003). Likewise, the risk of non-contact injuries was significantly lower when there was less water evaporation and higher rain fall (Orchard, Seward, McGiven, & Hood, 1999). Therefore, surface traction and the condition of natural turf surfaces are a major factor influencing injury risk when running is not performed in a straight line (Blazevich, 2004).

The friction/traction forces generated on playing surfaces describe the dissipative force that resists motion between two surfaces. Where friction is used to describe the resistance between two uniform, rigid surfaces, traction is used to distinguish friction-like shoe-surface interactions (Shorten, Hudson & Himmelsbach, 2003). Traction occurs during both running and turning. However, it is the increased medio-lateral forces that can be generated during lateral movements on surfaces with high friction/traction, which load the lateral ligaments and increase the risk of ankle sprain.

The traction forces generated at the interface between the shoe and surface should be such that the ankle ligaments are not subjected to excessive tension when the player performs a rapid change in direction (i.e. turning and cutting) (Durá, Hoyos, Martinez, & Lozano, 1999). When traction is low, the resistance to movement is limited and slipping is likely. Blazevich (2004) highlighted that although it may seem intuitive that surfaces of low traction which promote slipping would not be ideal, injury rates are generally lower on these surfaces. This is because slipping allows energy dissipation. When traction is too high, the foot sticks whilst momentum of the body continues forward. This excessive traction causes amplified horizontal forces, which encourages greater instability of the joint, and increasingly loads the associated structures (e.g. lateral ligaments, peroneal muscles, and other connective tissues). If the participant is not able to stabilise the joint motion, the foot rotates laterally, stressing the structures of the lower limb beyond safe levels, causing injury to the structures (Figure 2.14) (Blazevich, 2004; Luethi et al., 1986).

Despite the detrimental effect of excessive traction, the foot must experience some traction or else the participant will slip and fall. Therefore, with an optimal range of traction the participant does not fall, but does not experience excessive movements associated with injury. With optimal traction, the energy dissipation that the participant experiences is such that the velocity of the foot at braking is lowered, inducing a reduction in the velocity change and therefore a smaller change in momentum (Blazevich, 2004). Blazevich (2004) suggests that when the foot slides, the application of the horizontal GRF is slower compared to when the foot brakes instantly. These factors all combine to causes less lateral movement of the ankle, reducing the risk of ankle roll and placing the tissues of the ankle under less stress (Figure 2.19).

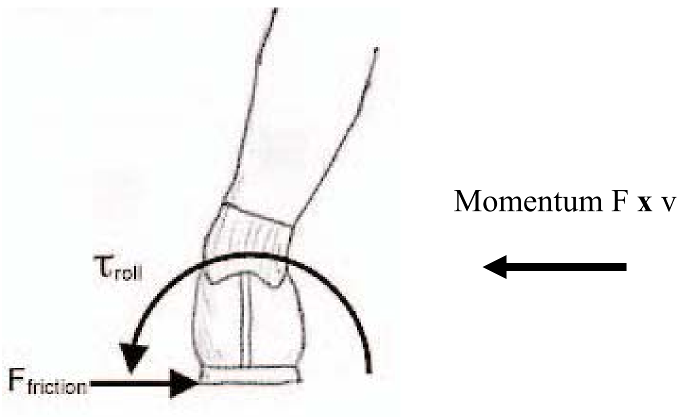


Figure 2.19. Interaction of momentum and friction on ankle movement, where F friction is the surface-footwear friction and T roll is ankle roll (Adapted from Blazevich, 2004).

To prevent high traction, it is important to understand how friction is generated between two uniform, rigid surfaces, and how it can be applied to a situation where there is an interaction between the shoe and surface. Friction occurs when any two surfaces touch each other. At the molecular level every surface has valleys and ridges, and at a given instant some of these will be touching. The response of these irregularities when sliding past one another depends on their respective material deformation properties (Barry & Milburn, 1999). However, Barry & Milburn, (1999) described that there is no "simple" theoretical model to calculate the friction between two given surfaces. Therefore, to explain the mechanism of traction provided by a playing surface is difficult, as classical laws of friction do not apply to footwear sliding on natural and artificial surfaces (Valiant, 1993 and de Lange & Winkelmolan, 1995 cited in Barry & Milburn, 1999; Brown, 1987; Shorten et al., 2003). These laws only apply to two dry, solid, and metallic surfaces sliding over each other (Barry & Milburn, 2005). Many, if not most materials used in footwear and playing surfaces, do not obey the classic laws of friction as there are often more than just two coefficients of friction (static and dynamic) (Brown, 1987).

The outsoles of soccer boots have various patterns or stud configurations that interact with either artificial turf or natural turf (Barry & Milburn, 1999). Because the physical interface between the surface and studded boot is different from normal sliding pairs used in friction studies, it is difficult to correctly determine the measurements of traction. However, while the mechanism of traction would be different, the reasons used to explain dry friction could provide the basis to explain the traction mechanisms associated with footwear-surface interactions (Barry & Milburn, 1999).

Soil structure consists of particles that are not strongly bonded together and can move relatively freely with respect to each other. Barry & Milburn, (1999) explained that surfaces are usually subjected to rain and the pore space between the soil particles can become partially filled with water and air depending upon the degree of pore space saturation. When a load is applied to the soil surface through the outsole and studs, the soil resists the applied loads by developing contact forces wherever they touch at their irregularities. As there are a large number of contacts within a soil mass, each contact causes the particles to respond by deforming in three ways: compressing, bending, and sliding. Deformation due to sliding is usually the most significant, and is nonlinear and irreversible, making the load-deformation behaviour of soil also nonlinear and irreversible (Lambe & Whitman, 1979 cited in Barry & Milburn, 1999). Because sliding between particles predominates, Barry & Milburn, (1999) suggested that the mechanisms used to explain dry friction can be applied to soils. They described that the external forces that cause sliding within soils are resisted by friction and bonding forces between the particles. However, the traction experienced by the soccer player is an interaction between the footwear and the surface. Thus consideration of the different types of surface-footwear interactions is required in order to quantify the traction provided to the soccer player.

2.4.2. Soccer specific footwear

2.4.2.1. Footwear traction.

In most sporting and physical activities, some form of footwear is worn (Willems, Witrouw, Delbaere, De Cock, & De Clerq, 2005). The design of the shoe is a paramount consideration to protect the athlete from injury. However, the design of the shoe can also affect the quality of performance to a much greater degree than most other protective devices (Cavanagh, 1985). As such, the protection of the athlete is often compromised, causing a number of situations of potential conflict where performance may dominate over protection (Cavanagh, 1985).

Figure 2.20. Example of a traditional soccer boot (Taken from Kays, 2007)

In soccer, the traditional footwear (Figure 2.20) worn by players was high sided (Lees & Nolan, 1998) to act as a restrictive layer to the ankle, supplementing the natural stiffness of both the talocural and subtalar joints (Johnson, Dowson, & Wright. 1976). This restriction aimed to reduce the foot-ankle complex movement because it improved the stiffness about the anterior-posterior axis through the ankle joint (Chomiak et al., 2000; Johnson et al. 1976). Johnson et al. (1976) found that the loads carried by the collateral ligaments will always be reduced by wearing a high type of boot as opposed to a low type. This is important as natural turf commonly provides an unstable platform on which to play, increasing the risk of excessive ankle inversion movement and potential injury. However, modern soccer boots are designed to be low cut, as soccer players favour a greater freedom of movement from the ankle joint in order for quicker and easier transition during multi-directional movements (Lees & Nolan, 1998). To provide this, the flexibility is maintained across the metatarsophalangeal joints (Chomiak et al., 2000) to enable a natural movement of the foot during propulsive phase, but the ankle support of the boot has been removed. This allows greater speeds of movement whilst also providing more rotation of the subtalar joint to dampen forces. However, it also results in less support at the ankle joint (Inkaar, 1994), causing a greater risk of injury during lateral movements compared to high sided boots.

To limit excessive inversion movements with low sided soccer boots, alternative design features have been incorporated into soccer boots. Because modern footwear has low sides, increased loads must be carried by the collateral ligaments compared to the traditional boot. Therefore, most soccer boots are low to the ground with very little mid-sole cushioning in order to enhance the performance of the athlete whilst trying to avoid the increasing rearfoot movement and the risk of injury (Morag, Amos & Bolt, 1994).

This risk of non-contact ankle injury is high in soccer, particularly during July, August, and September (Woods et al., 2003). One possible reason is that in soccer, the environmental conditions influence the choice of footwear worn by the player. Consequently, the aetiology of injury has been associated with an inappropriate choice of footwear (Inklaar, 1994; Wong & Hong, 2005). As the modern design features presented above are common in all soccer boots, and that the traditional soccer boot design is not used by modern soccer players, the failure to make a suitable footwear choice may relate to the design characteristics of the plantar sole, which influences the level of traction the participant experiences.

During lateral movements, traction forces are generated at the interface between the shoe and surface (Durá et al., 1999). As was discussed earlier with regards to playing surface, traction magnitude should not subject the ankle ligaments to excessive tension when the player is performing lateral movements (Durá et al., 1999). To achieve this, the combination of footwear design and surface must be appropriately chosen in order to permit adequate traction in such a way that the players can accelerate and decelerate, whilst still maintaining the balance, co-ordination and speed of movements (Durá et al., 1999).

The traction provided by soccer footwear is associated with the design of the plantar surface of the soccer boot. Across this region, multiple studs are located (Ihlenburg, 1999) which can differ in size, number, distribution and material (Daum, 1990 cited in González, Martinez, Montero, Alemany & Gámez, 2003). The primary aim of any studded footwear is to influence the level of traction provided during athletic movements (Lees & Nolan, 1998) and facilitate the anchoring of the shoe to the ground during running and changing directions (Daum, 1990 cited in González, et al., 2003). The design, construction and pattern of studs on the plantar sole of different soccer boots is therefore varied (Chomiak et al., 2000; Getz & Brannan, 2007) to suit the different surface conditions (Getz & Brannan, 2007) and provide optimal traction.

During the competitive period, soccer players tend to wear boots made with 6 removable screw-in studs (Santos et al., 2001). These studs can vary in size, but they are generally larger at the rear of the footwear (~ 13mm) than at the front (~ 11mm). The aim of these studs is to provide soccer players with traction during running and turning on soft (Santos et al. 2001), wet and unstable surfaces. An alternative to boots with screw-in studs, are those with moulded rubber soles. These boots tend to have 12 or more studs (Santos et al. 2001) that are shorter than the screw-in studs, and more greatly distributed across the plantar surface of both the heel and forefoot. Moulded soccer boots are worn when the surface becomes drier and more stable, but where moisture is still sufficient to cause slip. Soccer trainers have also been developed which do not use studs but are constructed with vulcanised rubber pimples across the plantar sole. Soccer trainers are designed for use on very hard surfaces or those constructed artificially. The exclusion of studs in this footwear type aims to enable stability to be maintained whilst combining with the surface to provide adequate traction.

Choosing the incorrect boot for a particular surface is likely to increase traction, which may cause the foot to 'stick', resulting in less horizontal slide and greater horizontal forces. This may create excessive ankle rotation during lateral movements and potentially cause overloading to the structures, and ultimately injury (Blazevich, 2004).

Therefore, because of the effect that the surface has on the choice of footwear, consideration into the correct boot is largely down to the perception of the surface condition, and an assessment of an acceptable level of injury risk that is presented by the surface conditions. Therefore, Livesay, Reda, & Nauman (2006) suggested the combination of shoe type and playing surface should be properly considered before use.

To assess the speculations that link the magnitude of traction to different footwear designs, soccer boots and their interaction with the surface have been tested using mechanical and biomechanical techniques. Nigg (1990) questioned the use of mechanical-based test procedures, although González et al. (2003) highlighted the difficulty for the athlete to perform “typical” movements. As such, mechanical devices have the advantage of creating conditions which provide repeatable measures of shoe-surface traction due to the variability in human performance (Clarke, Carré & Kirk, 2008).

Traction related to horizontal movements of the foot can be measured with linear traction tests. These involve dragging a loaded sled along the surface with a shoe attached, with the traction or horizontal force being measured throughout the test and maximum traction taken as a measure (Clarke et al., 2008). However, following high speed video observation of football movements, the initial movement of the shoe through the surfaces has been deemed more important. Clarke et al. (2008) used a device with a hydraulic ram to provide a constant vertical load to a studded plate and tested four different types of studs (2 rounded, Copa Mundial [short] and World Cup [long]; 2 uniquely manufactured long and short ellipse shaped studs) on a third generation artificial turf. They found that the geometry of the stud significantly influenced the mechanical traction. This was measured via the calculation of horizontal force when the stud had been translated by 10 mm. Specifically, they observed that the longer of the uniquely constructed ellipse stud had a reduced traction compared to the shorter ellipse stud, which was unexpected and was suggested to have been caused by the shorter studs being so shallow that the sole plate was touching the surface, acting as an additional source of traction. The authors subsequently suggest that the sole plate should be in contact with the surface during these tests so that to obtain comparative values of traction to those in real life (Clarke et al., 2008). On the other hand, the comparison of World Cup and Copa Mundial studs did not show similar trends, as the longer World Cup studs designed for softer surfaces produced more traction.

Coulomb law of friction can also be used to assess traction. A calculation of traction coefficient can be made by calculating the angle between the force acting normal to the surface and the force acting horizontally, and the friction coefficient is calculated in the same way as traction coefficient. (Shorten et al., 2003). Durá et al. (1999) evaluated the effectiveness of mechanical tests to replicate the traction that is generated by human participants. They used a German standard test (DIN18032-2), specific for use on sports surfaces, which involved the use of a sliding apparatus. This test consisted of a shaft with a diameter of 20 mm which is arranged in a frame, and the lower part which comprised of a threaded spindle. Durá et al. (1999) found that the traction coefficient calculated by the test apparatus did not yield similar values to the coefficient calculated during a biomechanical assessment of an athlete. It was explained that the differences in the coefficient of friction may result from adaptations and modification of the movements by the participant. When the frictional force is higher, participants spent more time in the braking phases of the movement and less time for starting (push off), thus producing the same effect in the mechanical impulse for each phase. The participants showed that extra time was gained for braking by creating significantly greater knee flexion, which could be interpreted as a protection mechanism, as participants try to maintain forces and torques within acceptable limits. When the tractional force is higher, the torque increases, but the effect of this is reduced by creating more time for braking via the added knee flexion. Durá et al. (1999) suggested that when the knee is flexed during the braking phase, an eccentric contraction of the quadriceps is produced and thus as the muscles became fatigued, this movement may produce other injuries. Durá et al. (1999) also concluded that a measure of turning performance is the speed of the movement. They reported that the time taken to perform a turn was not affected by the surface traction due to the compensation between the braking and starting phases (push off). Therefore, they recommend that low level traction is more appropriate to avoid injury without influencing performance.

To assess the effect of linear traction in soccer and the effect this has on the body, horizontal forces have been measured during sports-specific movements (accelerating from rest, inner and outer zigzag and turning movements) (Luethi et al., 1986; Stiles & Dixon, 2006; Gonzalez et al., 2003). This force indicates the level of resistance to horizontal movement and the loading that occurs to the lateral ankle. Likewise, Park et al (2005) used inverse dynamics to collect ankle joint moments, and hypothesized that an increased traction between shoe and surface influences joint moments and loads at the ankle. This hypothesis is consistent with the earlier suggestion that the increased traction would cause greater horizontal forces that must be resisted during turning and cutting. However, Park et al. (2005) compared the wearing of moulded and screw-in studded boots to running trainer, but found there were no significant differences between shoes eversion moments, suggesting that the risk of inversion injury did not increase. Park et al (2005) did however observe that in the frontal plane, the running shoe showed a trend towards higher ankle eversion moments than the three soccer studs. Furthermore, the increase in length of soccer studs (moulded: front 1cm/heel 1.2cm, studs: front 1.2cm/heel 1.5cm and blades: front 1.3cm/heel 1.6cm) associated with an increased ankle eversion moment. Thus, it could be speculated that an increase in foot height can cause greater instability, and increase the risk of injury. However, as these differences were all non-significant, it was proposed that the trends may be amplified during higher intensity movements. Likewise, if natural turf was used, then the traction may be sufficient to elicit significant differences. The measurement of both horizontal force and joint moments indicate the loading placed on the lateral ankle during turning and cutting movement and this can be used to indicate rearfoot inversion injury risk. As both horizontal force and joint moments require a force plate, the kinematic evaluation of peak rearfoot inversion may be a more flexible measurement of potential ankle loading when a force plate is unavailable, such as when testing on ‘real’ natural turf surfaces.

2.4.2.3. Footwear cushioning.

In soccer the choice of footwear has also been linked to the disproportionate number of overuse injuries during the preseason compared to the rest of the season (Woods et al., 2002). Injury can first be observed through increased pain experienced by the participant. Lafortune (1998) found that incidences of foot pain in soccer coincided with the positioning of studs. Lafortune (1998) noticed that foot pain occurred in the regions of the first and fifth metatarsal joints. Increased plantar pressures have been linked to this pain, although no direct association has been established. However, Martínez, Durá, Duenas, Gámez, Alemany and González (2004) identified that the regions of high pressure coincided with similar locations to both pain and the studs (i.e. the location of the first, and fifth metatarsals, and the heel). This pain associated with the position of the studs may result in response to the downwardly directed force generated during activity. This causes the studs to impact the ground directly under the structures where increased pain results. This is a major disadvantage with such design and as well as discomfort and pain, can result in bone bruising injuries and possible fractures occur in studded footwear sports such as soccer (Porter et al., 2005).

The configuration and level of penetration of the studs in the surface, as well as the sole plate stiffness, have been identified as factors that may elevate plantar foot pressure and increased foot pain (Shorten, 1998). Shorter, less numerous stud configurations which are less aggressive should be recommended on less adhering surfaces to decrease the risk of injury (Livesay et al., 2006; Torg & Quedenfeild, 1971). This increases the proportion of the stud that penetrates the turf, increasing the contact between the sole plate and the surface, and decreasing the pressure points across the foot (Kirk, Noble, Mitchell, Rolf, Haake, & Carré, 2007). Biomechanical data has been collected in response to the speculation that increased foot pain would occur with certain soccer boot designs. Heightened localised pressure was proposed to have been caused by a lack of material cushioning and may lead to foot damage (Coyles & Lake, 1999). However, Coyles and Lake (1999) found that there was no association between the magnitude of plantar pressure or loading rate, and the differences in stud material and location. Instead, they suggested that the similarities in pressure in their measurements must have been in response to similarity in stud number and location of the studs on the outsoles. Also, the combined influence of a sufficient stud number and appropriate outsole plate stiffness may adequately distribute the pressure across the forefoot, and this may lower the localised pressure experienced.

Dixon et al. (2008) used pressure insoles to assess the response of participants to three soccer specific footwear conditions. These included a screw-in and moulded studded boot, and a soccer trainer condition, which were tested on two natural surfaces that were distinctly different in their cushioning properties. It was found that the peak loading rate was greater when wearing the studded boot compared to the moulded boots whilst running on the harder surface, but not the softer alternative. It was concluded that the surface conditions are a fundamental consideration when the aim is to observe differences in the footwear conditions. As such, it may be speculated that the surface used by Coyles and Lake (1999) may not have been sufficiently hard to elicit changes in their kinetic measurements. Also, the findings of Dixon et al. (2008) indicate the role that the construction of the studs may have on the magnitude of loading rate during impact.

The response of the participant to the footwear condition may also depend on the identification of the correct movement patterns that are associated with the occurrence of injury (Stiles & Dixon, 2003; Queen et al., 2008). Queen et al. (2008) compared a soccer trainer (turf stud), blade and two moulded boots (firm ground and hard ground) during two types of cutting movement for male and female participants and observed significant differences, that were dependant on the movement performed and gender of the participant. Therefore, consideration must be given towards the movements and the participant gender when comparing different soccer footwear types as these factors can influence the conclusions of research papers.

During running and turning in soccer boots, it appears that greater loads are applied to the plantar surface of the forefoot and heel, depending on the boot design. Consequently, shoe design may play an important role in reducing loading rates applied to the forefoot and the underlying metatarsal bones (Orendurff, 2008; Martinez, Durá, Gaméz et al., 2004), as well as at the heel (Dixon et al. 2008). Therefore, although traction has been a key design goal in stud placement and shoe design, the available evidence suggests that an additional emphasis should be placed upon improving cushioning in the forefoot and heel of the soccer boot. The effects of which however, require investigation.

2.5. Footwear and surface changes

2.5.1. Footwear Cushioning

The level of cushioning provided to the athlete by the shoe has been suggested to be the most important function of footwear (Bates, 1985; Snel, Delleman, Heerkens & van Ingen Schenau, 1985). This is because footwear cushioning has been associated with overuse injury (Hennig, Valiant & Lui 1996) and the development of degenerative musculoskeletal diseases, due to repetitive loading (Radin et al., 1982; Voloshin et al., 1981; Voloshin & Wosk, 1982). To enable this attenuation, the material thickness of the mid-sole is commonly increased (Frederick, Clarke & Hamill, 1984), or a new material composition or design is developed. The premise behind these different cushioning techniques is that they attenuate the skeletal shock waves produced during locomotion. This occurs through passive attenuation of impact forces, and absorbing and dissipating the energy of a force as heat within the material. This can reduce the skeletal shock waves to a level that can be tolerated by the musculoskeletal system. By increasing the cushioning provided, the impact loading rate may also be lowered by delaying the onset and duration of the impact forces. In addition, by using greater cushioning, the contact area between the shoe and the foot can be increased. This increase in contact area may redistribute the plantar force beneath the foot to lower the peak pressures experienced at locations susceptible to injury.

To assess the protection that the footwear is providing, biomechanical and mechanical research has focused on rearfoot and forefoot shock absorption, as well as the degree of rearfoot control that is provided. These functions relate to the protection of the foot and the body from excessive forces and to the provision of adequate stability and control in the medio-lateral directions (Bates, 1985). Mechanical investigation of footwear is quick, but measures the forces that occur without the complex interaction of the participant. On the other hand, biomechanical testing is more laborious than mechanical testing, although it provides greater insight into the dynamic function of the shoe. However, the time and effort required to conduct biomechanical tests, makes them impractical for evaluating large numbers of shoes (Bates, 1985).

The cushioning provided by soccer boots has been compared to that given by running trainers. Santos et al (2001) observed that during a walking movement, maximum pressure was 35.0% greater in a studded boot compared to a running trainer. Similarly mean pressure was 27.6% higher in soccer football boots. Lees and Jones (1994) evaluated the shock experienced with different footwear by measuring the peak shank deceleration of soccer players. Whilst wearing standard running shoes, participants were asked to run on a treadmill, as well as on concrete and natural turf surfaces. On the natural turf, participants also wore a pair of soccer boots. Statistical comparison were made and it was shown that a significantly lower peak shank deceleration was observable between the grass and both the treadmill and concrete surface conditions whilst wearing trainers, but there were no significant difference between the treadmill and concrete. Significantly reduced peak shank deceleration was also observed when performing in the running shoe compared with soccer boot on the natural turf. However, there was no significant difference between peak shank deceleration on the concrete or treadmill in the running shoe compared to running on the natural turf in boots. As such, despite significant reductions in peak shank deceleration on the natural turf, wearing the less well-protected soccer boots increased the peak shank deceleration by about 10%. In this case the benefits of the softer surface were lost when using a boot that had no constructional mid-sole. The evidence would therefore suggest that the boot construction should include a mid-sole element, which would result in a reduction in impact severity.

Studies by Lees and Jones (1994) and Santos et al. (2001) showed the cushioning provided by the soccer boot is poor compared to running shoes. As such, even though the design of the plantar sole of the boot may influence the cushioning of the heel forces (Dixon et al., 2003), studded soccer boots should ideally have additional cushioning materials built in, but often they do not (Lees & Nolan, 1998). This would be particularly beneficial on the hard surfaces found during the summer, when injury rates are disproportionately high (Hawkins et al., 2001; Woods et al., 2003).

One way in which soccer players may experience increased cushioning on very hard surfaces is by wearing soccer trainers. By increasing the cushioning, it is thought that the forces occurring at impact, which are associated with many overuse injuries, would be reduced (Cavanagh & Lafortune, 1980; Clement et al., 1981; James et al., 1978; Munro et al., 1987). Mechanical tests consistently support this notion (Bates, 1985; Frederick, et al., 1984; Kaelin, Denoth, Stacoff, & Stussi, 1985), but biomechanical tests show no association between the size of peak impact forces and the level of mechanical cushioning provided by the footwear mid-sole (Clarke, Frederick & Copper, 1982; Cole, Nigg, Fick & Morlock, 1995; De Wit et al., 1995; Dixon & Stiles, 2003; Nigg, Bahlsen, Lüthi & Stokes, 1987; Praet & Lowerens, 2003; Snel, et al., 1985; Wright, Neptune, van den Bogert & Nigg, 1998). In fact some investigations have shown that a softer mid-sole results in a greater impact force (Hennig et al., 1996; Inkaar, 1994).

When soccer trainers have been used, Queen et al. (2008) found that participants experienced a significantly lower maximum force compared to a moulded and a bladed boot, which was most likely due to the additional cushioning provided by the mid-sole in the soccer trainer. Therefore, Queen et al. (2008) suggested that the dense stud configuration (vulcanised pimples) and additional cushioning provided by a soccer trainer may help prevent metatarsal injury and should be worn by athletes returning from metatarsal injury. However, Dixon et al. (2008) compared soccer trainers to screw-in and moulded soccer boots using peak impact force, and found that no significant differences between conditions were observed, which is consistent with previous cushioning investigations (Clarke et al., 1982; De Wit et al., 1995; Dixon & Stiles, 2003; Praet & Lowerens, 2003; Snel, et al., 1985; Wright et al., 1998).

Another method by which the cushioning provided by footwear can be compared is via the measurement of loading rate. De Wit et al. (1995) tested running shoes with different cushioning and reported that the rate of loading was significantly greater in the harder shoes than in a softer alternative. Likewise, Dixon (2008) published results that were consistent with DeWit et al., (1995), when comparing cushioned running shoes with a shoe with little cushioning (racing shoe). However, Nigg et al. (1987) failed to find any difference in loading rate between differently cushioned footwear. Likewise, Dixon et al. (2008) found that differences between soccer boots and soccer trainers were not found when measuring impact loading.

Shorten (2002) described that the forces occurring at impact were unrepresentative of those forces occurring at the heel. Shorten observed that impact forces occurred during the first 50 ms and were composed of the summation of heel, mid-foot and at times, forefoot force. Because of this, heel forces have been measured to represent the force that propagates the lower extremity (Clinghan et al. 2008; Voloshin & Wosk. 1982). Dixon (2008) hypothesised that in footwear with a less cushioned heel, a significant increase in peak forces and peak loading rates at the heel would be observed. Despite this hypothesis, peak forces at the heel did not differ between the footwear. However, Dixon (2008) did observe significant differences between the cushioned and un-cushioned running shoes for the measurement of peak rate of loading at the heel. Dixon reported that the footwear with minimal cushioning had higher rates of loading than all but one of the shoes provided with cushioning. Dixon also emphasised the importance of considering loading across different regions of the plantar foot when studying different footwear types. Significantly lower peak forces at the mid-foot were observed with one of the cushioned shoes compared to the shoe with no cushioning. These findings are consistent with models of contact between non-conforming elastic bodies and, in contrast to GRF measures, appear to suggest that shoe cushioning does influence the way the body is loaded (Shorten, 2002).

Another way in which regional load can be measured is by calculating peak plantar pressures. Peak pressures take into consideration both the force magnitude and the area which it covers. Peak plantar pressure measurements have been used for biomechanical assessment of cushioning and have been able to distinguish differences between footwear with different mid-sole cushioning (Bus, Ulbrecht, & Cavanagh, 2004; Hennig & Milani, 1995). The results of these studies could relate to greater relative contact so that the force is applied to a larger area, thereby lowering the pressure (Burnfield, Few, Mohamed & Perry, 2004). The greater contact area may also provide a mechanism by which the load is redistributed away from bony prominences to more peripheral regions to reduce pressure at key locations (Bus et al. 2004; Lees & Nolan, 1998).

Utilising these measurements, it may be possible to establish significant differences between soccer boots with soccer trainers. Therefore, in the study of footwear in soccer, consideration should be given to using a range of biomechanical measurements.

2.5.2. Additional footwear cushioning techniques.

2.5.2.1. Cushioning insoles.

Although soccer trainers may increase cushioning during certain movements, most soccer players commonly wear the same soccer boots all year. To improve footwear with poor force attenuation properties, cushioning insoles (Dixon et al., 2003; House et al., 2002; Windle et al., 1999) and heel inserts have been used (MacLellan & Vyvyan, 1981; Light & MacLellan, 1977; Nistor, 1981). Insoles and heel inserts both refer to material that can be placed in to the shoe, and are often terms used interchangeably. However, for the purpose of this thesis, the term cushioning insole refers to a full length insole that covers the entire plantar foot (Figure 2.21), whereas an insert refers to a device that is positioned only at the heel (Figure 2.22).

Figure 2.21. An example of the cushioning insole

Figure 2.22. An example of a heel insert

Cushioning insoles have been used in athletic footwear to reduce the forces associated with foot pain (Shorten, 1998). The use of visco-elastic insoles have also been reported to produce a significant shift in the location of pain from the back to the lower extremities (Tooms et al., 1987), which may suggest that the forces experienced could be either attenuated or relocated, which effectively lowers the concentration of force/pressure in one location. Gardener, Dziados, Jones, Brundage, Harris, Sullivan & Gill (1988) highlighted that many citations can be found which recommend the use of shock absorbent insoles based on a review of clinical cases.

In a prospective study of 90 US Coast Guard recruits, Smith, Walter & Bailey (1985) found that cushioned insoles were successful in reducing "shock impact and shearing injuries". However, the impact and shearing injuries referred to were foot bruises, calluses, and blisters, and not stress fractures or stress reactions (Gardener et al, 1988). Gardener et al. (1988) observed that the risk of stress fracture did not change between two cushioning insoles constructed from a polymer Sorbothane insole and a standard mesh insole. This finding is consistent with the biomechanical study conducted by Nigg, Herzog and Read (1988) who found that visco-elastic insoles did not attenuate peak impact forces when running at 4 m/s in running shoes. However, studies of cushioning insoles have shown that much variation occurs between insole and non-insole conditions, as the construction and material of the insole are important considerations, as is the biomechanical measurements used. Likewise, the age of the insole has also been shown to influence the research finding. Therefore, the response of participants may have been different if the insoles used had fully considered the effect of all of these factors, with which they did not.

Rööser, Ekblacdh and Lidgren (1988) aimed to understand the effect of insole thickness on the loading of the lower extremity. They observed that an inlay material of 6 mm did not significantly reduce the tibial acceleration produced at heel strike, whereas a 20 mm insole did, which suggests a reduced risk of impact injury. Windle et al. (1999) also investigated the shock attenuation characteristics of three different insoles worn by military recruits along with a control condition (no insole) during running and marching using in-shoe pressures. The insoles included a visco-elastic polymetric insole (Cambion®), polymetric foam insole (PPT®), a military issue Saran insole and a Sorbothane® insole. All insoles significantly reduced peak pressures during heel strike and forefoot loading compared to the control. The authors found that the peak pressures were reduced most with the Sorbothane® insole, followed by the visco-elastic polymetric insole, polymetric foam insole and the Saran insole. The Sorbothane® insole resulted in a significantly lower peak plantar pressure at heel strike compared to the other conditions. During forefoot loading the peak pressure attenuation of all four insoles was similar, except for significantly lower peak pressures in the Sorbothane® insole compared to the visco-elastic polymetric insole.

House et al. (2002) compared four insoles that had experienced mechanical degradation equivalent to 100-130 km of running. The peak pressures were then collected whilst wearing a combat assault boot (CAB) using an in-shoe pressure measuring system. Insole 1 (Sorbothane®) was a moulded polyurethane foam with a 1 mm layer of polyurethane elastomer inserted at the heel and forefoot. Insole 2 (Sorbothane®) was constructed of a 3 mm polyurethane material. Insole 3 was foam shaped attached to a high-density ethyl vinyl acetate (EVA) footbed (Poron) and insole 4 consisted of a 3 mm base of coarse weave plastic with a top sheet of nylon non-woven fabric (Saran). Peak pressures were compared between each insole, as well as a control. The main finding presented by House et al. (2002) was that the insoles did not lose their ability to attenuate peak pressures after degradation and that on average the peak pressures at the heel and forefoot were reduced by 37% and 24% respectively, compared to the control.
Dixon et al., (2003) collected data during shod running with the same four insole

conditions as House et al. (2002), but where House et al. measured peak pressures, Dixon et al. analysed peak impact forces and loading rates. The insoles were tested before and after mechanical degradation equivalent to 100-130 km, which resulted in an increase in mechanical stiffness and a decrease in ability to reduce mechanical impacts for all test insoles. However, no significant differences were found for peak impact force between conditions, which is consistent with a plethora of previous cushioning related research (Clarke, Frederick, & Cooper, 1982; Dixon & Stiles, 2003; De Wit, et al., 1995; Praet & Lowerens, 2003; Snel, et al., 1985; Wright, et al., 1998). However, Dixon et al. (2003) did find significant differences when analysing the peak rate of loading for each insole condition. When participants wore insole 3 (Poron) without degradation, significantly lower loading rate was observed compared to a control. The ability of this insole type to reduce peak rate of loading during running was maintained after mechanical degradation. A significantly lower peak impact loading rate was also observed for insole 1 (Sorbothane) when degraded, insole 2 (Sorbothane) when new and degraded, and insole 3 (Poron) when new and degraded, compared to insole 4 (Saran) when degraded.

Eils & Streyl (2005) examined the effect of aged insole placed within a soccer boot on the loading of the plantar foot. Using a new shoe with a new insole, in-boot pressures were collected. The boots was then worn for a year, and then re-tested. The insole was then replaced and the old shoe with a new insole was tested. The comparison between three different aging conditions showed no significant differences for the measurement of relative loading for any of the 10 different foot areas measured. However, as there was no controlled degradation of the insole and shoe, the amount of wear may have been insufficient for differences to be observable.

In summary, the results of these studies suggest that the response to cushioning insoles is not clear and much variation occurs due to the type and age of the insole worn. Dixon et al. (2003) showed that cushioning insoles may be useful in the reduction of impact loading, although the cushioning provided by the insole is dependent on the age and the material which is used. Windle et al. (1999) also suggest that the response of a participant to a cushioning insole may change depending on the footwear worn. For example, when cushioning insoles are placed into running shoes that already provide inherently good levels of cushioning, the response may be different than if placed inside footwear with limited or no cushioning. It should therefore not be assumed that the result of a cushioning insole study would be replicated if different footwear types were tested (Windle et al. 1999). Likewise, differences may be shown if along with running, other movements are performed.

2.5.2.2. Heel inserts.

As well as full length cushioning insoles, commercially available visco-elastic heel inserts have been used in stiff mid-sole shoes as part of a comprehensive treatment program to treat overuse injuries (Johanson, Cooksey, Hillier, Kobbeman, & Stambaugh, 2006). The use of heel inserts of between 6 mm (Leach, James & Wasilewski, 1981) to 15 mm (Smart, Taunton, & Clement, 1980) has been recommended in order to lower heel and ankle pain (MacLellan & Vyvyan, 1981; Ramanathan, John, Arnold, Cochrane, Abboud, 2008). These inserts also reduce the soreness in the calf, back and Achilles tendon in soccer referees (Faunø et al., 1993), and soccer players during preseason (MacLellan, 1984), as well as lowering the incidence of injury in non-injured athletes (MacLellan & Vyvyan, 1981; DeMaio, Paine, & Drez, 1995; Mazzone McCue, 2002; Leach, James, & Wasilewski, 1981; Maclellan & Vyvyan, 1981). Smaller heel inserts of 2.5 mm have also been shown to benefit athletes with severely damaged Achilles tendons, and may act as part of a suitable alternative to surgical operations. Nistor (1981) compared two groups in which one group experienced Achilles tendon surgery and the other was a non-surgical group that had a below the knee cast applied for four weeks in gravity aided plantar flexion. This was then followed by a reduction in the equinus angulation (this is a fixation of the foot, or part of a foot, in a plantar flexed position) and the application of another cast. These casts were worn for a total of 8 weeks. Once the second cast was removed, a 2.5 cm heel insert was used for four weeks or until 10° of dorsi-flexion was achievable. This treatment was shown to match the surgery alternative, but complaints due to the treatment were less, and no hospital stay was needed. However, the evidence regarding the use of heel inserts interventions is not completely indisputable. Lowden, Bader and Mowat (1984) compared two types of heel inserts (Sorbothane and Molefoam) with a control group. The control group showed a more rapid improvement in swelling and tenderness, pain and activity levels, leaving the reported benefits of the heel inserts unsubstantiated. However, these authors concluded that because only 33 participants were used, which were divided in to one of three groups, the sample size was small and this could have concealed a treatment benefit of the heel insert.

Despite the literature generally advocating the use of heel inserts to reduce overuse injury, specifically to the Achilles tendon, the mechanism behind this success is unclear. Lower extremity biomechanics such as tibial acceleration have been shown to be influenced by othotic interventions (Bojsen-Moller, 1983; Light & MacLellan, 1977; Light et al., 1979), as has GRF and peak loading rate (Lees & McCullen, 1984), and peak rearfoot movement (Stacoff & Kaelin, 1983). However, Dixon (1996) indicated that there is missing scientific evidence correlating the reduced incidence of Achilles tendon injury with a heel insert using biomechanical variables. Dixon (1996) highlighted that the results of clinical studies such as those presented in the current chapter describe the benefits of heel inserts interventions performed by experienced practitioners whose knowledge in treating Achilles tendon injury cannot be ignored. Dixon (1996) suggested that further biomechanical studies of heel inserts interventions are required to improve our understanding and aid future development of methods to reduce injury.

Heel inserts are used to alter lower extremity joint kinematics and temporal variables during weight bearing activities to reduce stress on affected tissues (Johansson et al., 2006). Light and MacLellan, (1977) measured the in-bone impact shock on the tibia during heel strike and modelled the distortions produced by this shock which is transferred from the bone to the soft tissue. It was demonstrated that substantial traction, shear and an overswing or “whipping” of the tendon can take place which results from this transfer of shock, but Light and MacLellan found that the impact shock measured was significantly reduced with the inclusion of a visco-elastic heel inserts. They suggested that by reducing the shock wave, soft tissue damage to structures such as the Achilles tendon could be reduced.

MacLellan and Vyvyan (1981) speculated that a reduction in injury and heel pain would occur with commercially available heel inserts because the visco-elastic material would attenuate the forces that occur at impact. MacLellan & Vyvyan, (1981) supported this speculation by observing a reduced incidence of heel and Achilles tendon pain with commercially available heel inserts. However, despite the reduced incidence of Achilles tendon injury with the use of the heel inserts, no measurements of tibial acceleration were taken to support the suggestion that the heel inserts had lowered the incidence of injury through increased impact attenuation. Dixon and Stiles (2002) used peak impact force as an in-direct measurement of the impact shock. The authors found that no significant difference was observed for either peak impact loading or the magnitude of vertical force when a 6 mm or 12 mm heel insert was applied. Likewise, Amos, Bolt and Morag (2004) tested the effect of 6 mm EVA heel inserts on reducing the heel pressures in children and found that the heel pressure was not significantly reduced.

Another possible explanation for the success of the intervention is that the inclusion of a heel insert enables the heel of the athlete to be raised relative to the forefoot (Clement et al., 1984; Leach, et al., 1981). During mid-stance this reduces the maximum ankle dorsi-flexion angle (Clement et al., 1984), influencing the moment arm length (Dixon & Kerwin, 1998; 2002) and the magnitude of tendon lengthening. During the cyclic loading found during running, multiple repetitions of dorsi-flexion movement occur. The use of heel inserts may reduce the maximum dorsi-flexion that occurs during running and place less accumulated eccentric strain on the Achilles tendon. This is because as dorsi-flexion magnitude is reduced, the muscle group has to perform less control.

Kinematic measurements of ankle movement have been reported for a variety of heel insert sizes. Ramanathan et al. (2008) found that the kinematics of participants changed with the use of an “off the shelf” insert whilst walking. Johansson et al. (2006) also found that during a walking task, the use of a 9 mm heel insert resulted in the greatest increase in ankle dorsi-flexion excursion compared to a 6 mm heel insert and walking shoes alone. It is possible that heel inserts position the ankle in a more plantar flexed position at foot flat, which may allow a greater range of ankle dorsi-flexion excursion between foot flat and heel off in participants with tight gastroncemius muscles. This is in contrast to the hypothesis regarding reduced dorsi-flexion with heel insert use. Also, the athletic shoes used in the study had a heel wedge built into the shoe, which may affect gait differently than heel insert secured to bare feet or in other footwear types (Johansson et al., 2006). As such, the condition of the gastroncemius muscle and participant footwear may influence the response to the heel insert.

Dixon and Kerwin (1999) measured the effect of a heel insert during running. It was reported that peak dorsi-flexion was significantly reduced when a 15 mm heel insert was used compared to a control condition. However, when a smaller 7.5 mm heel insert was compared, no differences were observed. Likewise, Dixon and Stiles (2002) measured peak dorsi-flexion angle whilst wearing footwear with a 6 and 12 mm heel insert and found that peak dorsi-flexion was significantly reduced with the 12 mm heel insert but not the 6 mm.

The assessment of overuse injury using lower limb kinematics is relatively quick and simple. However, lower limb kinematics does not indicate the internal forces produced by the gastroncemius muscles that load the Achilles tendon whilst performing relevant movements. Forces produced by muscles are quite sensitive to changes in lever length. Therefore, the response of the athlete to heel inserts may not always be observable with external measurements. Instead, the calculation of peak plantar flexion moments has been used to estimate the net force occurring to control dorsi-flexion. Using this measurement, Reinschmidt and Nigg (1995) investigated the effect of varying heel heights on the participants' plantar flexion moment. These authors found that the group mean of maximum plantar flexion moment and its time of occurrence were not significantly affected by heel height.

Although the measurement of plantar flexion moment can indicate the net forces during dorsi-flexion, the magnitude is not an estimate of Achilles tendon force. Dixon and Kerwin (1998) analysed the factors that contributed to the observed changes in maximum Achilles tendon force. The moment arm of the Achilles tendon about the ankle joint centre was highlighted as a potential factor that would influence Achilles tendon loading. Therefore, Dixon and Kerwin (1998) used plantar flexion moment occurring in the sagittal plane, alongside a calculation of the moment arm distance from the ankle joint centre to the line of the Achilles tendon (represented by 2 markers on the rear leg), to calculate Achilles tendon force. The authors found that the response to the heel insert differed depending on the running style of the individual. A rearfoot and mid-foot striker demonstrated significantly increased maximum Achilles tendon force whilst using a heel insert, whereas a forefoot striker exhibited no difference. The different response to this intervention between subjects of different running styles highlighted the importance of classifying the subjects based on their running style. Dixon and Kerwin (2002) went on to investigate the influence that three insert conditions (7.5 mm and 15 mm heel inserts, and control) would have on the peak Achilles tendon forces and loading rate for a group of barefooted heel toe runners. The authors found that despite group reductions, the magnitude of the peak Achilles tendon forces and the peak rate of loading were not significantly different between conditions. However, Dixon and Kerwin (2002) did find that the use of the 15 mm heel insert significantly reduced average Achilles tendon loading rate, suggesting that the average rate of loading may be a more important measure when assessing the mechanism behind Achilles tendon injury risk.

McGuigan, Dorey & Lichtwark (2007) evaluated the effect of heel inserts on the Achilles tendon during walking and running. These authors strapped a 12 mm EVA foam heel insert to the plantar foot to change its orientation, and used simultaneous kinematic and ultrasound measurements to calculate the Achilles tendon strain during running and walking. To provide the ultrasound data during walking and running, an ultrasound probe was taped to the limb at the gastroncemius muscle-tendon junction to image the displacement of the junction. McGuigan et al. (2007) then compared the Achilles tendon stain under the 12 mm condition to the strain observed when an 18 mm heel insert was used as well as to a control condition. However, the measurements revealed no significant differences. McGuigan et al. (2007) explained that the lack of significant difference in Achilles tendon strain is likely due to the bi-articular nature of the gastroncemius muscle and the relative timings of the ankle dorsi-flexion and knee flexion during stance. This would influence the relative change in Achilles tendon length. However, Dixon and Kerwin (1999) also estimated the strain placed on the Achilles tendon. Using the findings of Grieve et al. (1978), Dixon and Kerwin (1999) determined the change in the peak gastroncemius muscle tendon length to indicate a change in Achilles tendon length. It was observed that the change in the Achilles tendon length was significantly less with the use of a 7.5 and 15 mm heel lift compared to a zero heel lift condition, where the 15 mm heel lift resulted in the largest decrease in Achilles tendon lengthening. Dixon and Stiles (2002) also observed significant difference to peak Achilles tendon strain when using a 12 mm, but not a 6 mm, heel insert placed inside a shoe.

The findings related to participants' kinematic, kinetic, joint moment and Achilles tendon loading with the placement of a heel insert into the footwear has indicated very few findings. Therefore, the mechanisms behind the success of the heel insert to reduce Achilles tendon injury remains unclear. One proposed reason for such findings may relate to methodological limitations. For example, Dixon and Kerwin (1998; 2002) used two-dimensional sagittal plane kinematics to calculate the plantar-flexion moments occurring during running. However, as the foot is able to move three dimensionally, this could influence the accuracy of the plantar flexion measurement. This may also influence the calculation of the moment arm between the ankle joint centre and the line of the Achilles tendon. Therefore, if a three dimensional data is used, improved calculation accuracy is gained, and greater insight into the injury protecting benefits of the insert is provided. Also, Dixon and Kerwin (1998; 2002) used barefooted runners to assess the effect of the heel inserts, which may affect the response to the heel insert compared to shod running (Johansson et al., 2006). The response to the intervention may also depend on the individual. Runners are biological organisms with anatomical and functional differences, and as Bates et al. (1983) highlighted, there is some variability between performers and within the same performer over repeated trials. As such, there could be large variability in kinematic measurement between and within subjects. The individual difference in the anatomical structure and biomechanical performance characteristics may also cause large standard deviations within the group means and therefore influence the ability to observe significant group differences. It may therefore be more appropriate to present data for individual subjects, eliminating the possibility that inter-individual differences may obscure the behaviour of participants. In addressing these limitations, it may be possible to highlight significant differences between the heel insert conditions and thus these can be used to highlight the mechanisms behind injury reduction, which previous investigation has failed to do. However, one point also worth considering is that the lack of significant differences may have been caused because the difference between the footwear conditions is not significant. Therefore, either the injury studies are misguiding, or the success of the heel insert inclusion is based on other currently unknown measurements.

Additional problems may also arise if heel inserts/insoles is placed inside a soccer boot. Crosbie and Ko (2000) have shown that when stepping over a kerb under natural environmental conditions out of doors, accurate planning obstacle avoidance occurs some distance before the obstacle is reached. A similar mechanism may operate for walking, running, and changing direction on smooth surfaces. If inversion, plantar flexion, and rotation angles during the swing phase are set relative to the perceived angles from the previous stance phase, then proprioceptive error could accumulate to the point where the position of the foot and ankle at the start of the stance results in excessive motion and injury (Waddington & Adams, 2003). Waddington and Adams (2003) found that when wearing soccer boots the ability to avoid obstacles was less when no soccer boots are worn. Consequently, it can be speculated that by placing even greater material between the foot and the ground, movement discrimination would be worse, thereby increasing the risk of placing the foot in an inappropriate position. In such cases, the difference between the perceived distance between the foot and the ground and the actual distance may cause the foot to contact the ground incorrectly and cause injury.

The effect of the heel inserts during turning may also increase the instability of the rearfoot by placing the foot into a more plantar flexed position. In doing so, the loading of the ankle ligaments are increased (Fujji et al., 2005). Further still, this additional material supplied by both the inserts and insoles may compromise the design characteristic of the boot that supplies medio-lateral stability (e.g. the stiff sole plate). The greater visco-elastic material may be more easily compressed on the lateral edge to further increase the maximum inversion magnitude. Because of these factors, there may be an increased risk of acute injury, particularly during lateral movements in soccer which may act as another factor by which lateral ankle injuries can be sustained. As such, despite the recommendations that may be given regarding the use of heel inserts and insoles in soccer, the use of such devices may increase the risk of injury to other structures. Therefore, although heel inserts have been worn by soccer players and referees (Faunø et al., 1993; MacLellan, 1984), the recommendations for using heel inserts and cushioning insoles may be different if these factors are influential in the risk of sustaining additional injuries. This has not been previously investigated.

2.5.3 Artificial turf playing surfaces in soccer

2.5.3.1. Traditional artificial turf.

Natural playing surfaces are the preferred choice for the majority of soccer players (Martinéz, Durá, Gaméz, et al., 2004). However, the quality of natural turf surfaces is influenced by environmental conditions, which are not always favourable for use in soccer. Some climates can discourage grass growth, hampering the development of the sport in those countries (FIFA, 2001). In temperate climates found in Britain, France and Germany, playing surfaces are regularly exposed to excessive precipitation during the winter months, which can cause quick degeneration of the surface quality.

The design of the modern stadia is “closed”, meaning they have high sides with partially enclosed roofing, making them public friendly (Veenbrink, 2002). This design characteristic can make natural turf surfaces expensive and difficult to maintain. This is because the surface can experience a reduction in the sunlight and wind necessary for turf growth and removal of surface moisture, respectively. As a consequence, when a natural turf surface experiences extreme weather conditions and is frequently used throughout a season, the quality of the surface can be substantially reduced, influencing both the quality of the player's performance and the risk of injury. Consequently, the turf must be replaced a number of times each year (Jowett, 2004; Veenbrink, 2002). The roofing can also cover the entire stadium. This aims to prevent extreme environmental conditions influencing the surface characteristic, but can prevent sufficient turf growth (FIFA, 2001). Another problem with natural turf is that the quality of the surface changes throughout the year due to climatic differences. During the summer months the playing surface can dry out and become hard due to increased temperature and reduced precipitation. This hard surface can influence the surface cushioning and traction characteristics, making it very uncomfortable to play on and increase risk of injury (Hawkins et al. 2001; Woods et al., 2002).

In response to the limitations of natural turf, artificial surfaces were developed (FIFA, 2001). These surfaces aimed to withstand the harsh and extreme environmental conditions whilst lessening the influence of these adverse weather conditions on surface playing ability and reduce the cost of maintenance (Kolitzus, 1984; Nigg & Yeadon, 1987). Artificial turf surfaces also aim to replicate the playing characteristics of natural surfaces (Kolitzus, 1984; Nigg & Yeadon, 1987), whilst increasing player comfort and safety, being low in maintenance, multiuse for different sports, having extended usage and being useable all year round (FIFA, 2001).

The original artificial surface used in soccer was sand based and during the 1970s and 80s some of Britain's elite soccer clubs use these surfaces for competitive matches. This first generation of artificial turf surfaces was made from short, tightly matted carpets that were sand filled and lay over rubber shock pads and an engineered base (The FA, 2005). However, while aspects of athletic and gymnastic performances have improved with the use of artificial surfaces (Stiles & Dixon, 2007) the early generations of artificial surfaces were not specifically designed for use in soccer (FIFA, 2001). Although they could sustain much higher levels of use than natural grass, the playing qualities of sand filled carpets were not particularly good for soccer (The Football Association, 2005). The inherent problem with the first generation artificial surface was the accentuated and uneven bounce of the soccer ball, as well as an increased risk of severe abrasion. The first generation artificial surface was also relatively hard compared to natural turf and the use of the relatively stiff artificial surface has and been associated with an increased prevalence of injury compared with natural turf surfaces (Arnason et al., 1996; Nigg, Cole, & Stefanyshyn, 2003; Nigg & Yeadon, 1987; Ramirez, Schaffer, Shen, Kashani, & Kraus, 2006). Some possible causes for this increased risk of injury include an increased force magnitude at impact, altered joint movement patterns, and differing resistance to sliding between the shoe and surface (Nigg, Frederick, Hawes, & Luethi 1986; Stucke, Baudzus, & Baumann, 1984). However, ridicule from fans and players alike meant that soccer failed to embrace this surface (FIFA, 2001) and most soccer clubs reverted back to natural turf surfaces.

Biomechanical research

Biomechanical research has been performed on a variety of diverse and mechanically distinct artificial surfaces (Dixon & Stiles, 2003; Stiles & Dixon, 2006; Stiles & Dixon, 2007). Shorten and Himmelsbach (1999) measured impact shock through tibial accelerations during controlled landing on to natural and artificial turf. It was observed that a reduced impact shock occurred on the natural turf surface which was consistent with a mechanical test and with subjective reports of perceived impact severity. In contrast, studies that use peak impact forces to estimate the shock that occurs internally, typically report a maintained peak impact force magnitude across surfaces (Bobbert et al., 1992; Clarke, Frederick & Copper, 1982; Dixon, Collop & Batt, 2000; Dixon & Stiles, 2003; Stiles & Dixon, 2007; Nigg & Yeadon, 1987). However, most of these studies have used a running task. Stiles and Dixon (2006) found when using a dynamic and tennis specific movement, impact forces could be used to distinquish between surface conditions, yet in the opposite direction that would be expected. They observed that the tennis surfaces with the highest mechanical cushioning resulted in the highest vertical force magnitude.

Due to the link between injury risk and loading rate (Hreljac, 2004; Hreljac et al., 2000; Radin et al., 1991), the measurement of average and peak loading rate has also been recorded. It has been shown that with a decrease in the cushioning of artificial turf, significantly greater loading is experienced (Dixon et al., 2000; Stiles & Dixon, 2007). Stiles and Dixon (2007) observed that significantly lower peak loading rates were evident on two foam surfaces compared to a harder acrylic and rubber surface. Stiles and Dixon (2007) also found that the time of peak impact force was significantly later in the foam condition than in the acrylic and rubber condition. This smoothing of the impact peak created distinctly different initial force characteristics, and therefore the peak rates of loading were systematically and significantly lower on the two foam conditions than on the baseline acrylic condition (Stiles & Dixon 2007). Along with the use of peak vertical force and rate of loading, Stiles and Dixon (2007) measured the peak pressures occurring in response to mechanically distinct surfaces. It was found that peak heel pressures showed systematic and significant reductions with increased surface mechanical cushioning. They concluded that compared with the acrylic and rubber, the greater deformation potential provided by the foam condition, increased the contact area between the shoes sole and surface. The provision of additional contact area may allow a greater force distribution over the heel. This may result from greater deformation of foam causing a reduction of the peak pressures experienced by the participant (Hennig et al., 1996; LeVeau, 1992; Stiles & Dixon, 2007). This evidence suggests that if careful selection of the biomechanical measurements and movement patterns are performed, significant differences between natural and artificial surfaces can be observed.

2.5.3.2. Third generation artificial turf.

Undeterred with the rejection of the first generation surface, manufacturers designed new surfaces (Jowett, 2004), which are termed "infill" systems or the third generation artificial turf. These surfaces are comprised of several layers, and are often divided into the 'foundation' and the 'surface system' (Fleming, Anderson & Ansarifar, 2008). The 'foundation' generally comprises of an engineered sequence of porous bound macadam, crushed rock and natural soil. This provides a solid platform which stays porous, stable and 'flat' for at least 25 years (Fleming et al., 2008). The 'surface system' comprises of the carpet (either filled or unfilled), overlying a shock pad of which there are various types (Fleming et al., 2008). The carpet consists of a horizontal backing supporting numerous vertical nylon or polypropylene fibres. These vertical fibres (pile) are much longer and more thinly spaced than the traditional synthetic turf (first generation), and unlike the original artificial turf surfaces, can be filled with varying types of granulated material. This is typically a sand and rubber infill media (Popke, 2002; The FA, 2001). Each of the components of the system interacts to provide an aspect of performance required by the surface (Fleming, et al., 2008). As a consequence, the third generation turf surfaces are said to be more soccer specific and mirror the characteristics of natural turf closely (Ekstrand, Timpka & Hägglund 2006; FIFA, 2001; Meyers & Barnhill, 2004; Popke, 2002; Veenbrink, 2002), whilst also maintaining a consistent quality throughout the season (Veenbrink, 2002).

Comparisons of injury rates between third generation artificial turf and natural turf surfaces have shown that the rates and severity of injury are similar for all injury types during matches and training per 1000 hours of exposure (Ekstrand, Timpka, & Häglund, 2006; Fuller Dick, Corlette, & Schmalz, 2007; Steffen, Andersen, & Bahr, 2007). However, these comparisons are missing a number of key factors. Firstly, the nature of overuse injuries means that they occur over a sustained period of time. However, as injury surveillance studies typically follow one group and record the surface on which the injury occurred, the micro trauma development may have taken place on one surface used for training, only for the injury to be observed whilst playing on another surface. In this situation, the surface that the injury occurs on may be recorded incorrectly. Further still, the rate of overuse injury is typically reported as a yearly total. No comparison has been made between the two surface conditions at various times of the year. Due to the injury rates on natural turf changing throughout the year (Hawkins et al., 2001; Woods et al., 2002), it could be speculated that the risk of injuries on the artificial turf could follow this trend. On the other hand, as the third generation surface is designed to be more consistent over time, the surface may offer potential reductions in injury rates during some periods whilst increasing the risk at others. As a result, the total annual injury rate may be similar but the relative risk of injury may change. Therefore, if it is assumed that the artificial surface will stay consistent in both cushioning and traction properties over time, it would be expected that differences in the biomechanical data would be observed when compared to a natural turf at different times of year. When the climate is wet and cold, natural surfaces can become sodden which can provide an effective method of shock absorption compared to third generation turf, which is perceived to be less cushioned than natural turf (Martínez, Durá, Gaméz et al., 2004). Therefore, greater impact loading and pressures may be experienced on the artificial surface. Conversely, as the climate becomes warmer, the cushioning of natural turf is reduced. As artificial turf is provided with a shock pad and rubber the peak forces, and impact loading may be reduced on this artificial surface compared to natural turf surfaces. Further still, as the studs may more easily enter the artificial surface compared to a hard natural turf, the interaction between the stud number and height will influence the contact area between the foot and the ground. Thus the plantar pressures and pressure loading rates experienced in the different footwear may be increased when running on a third generation artificial surface compared to a soft natural turf, or decreased on the artificial surface when the natural turf is hard. Likewise, as the interaction between the boot and the surface determines the level of traction given to the soccer player, the difference in surfaces at different times of the year may influence the risk of acute injuries.

The response of the soccer player to the artificial surface in comparison to natural turf may also depend on the construction of the surface. There are many aspects of the design of the artificial turf that can change the properties of the third generation artificial surface (Fleming et al., 2008). The most important of these are rubber size, binder content, shock pad layer thickness, and bulk density. However, the effect of these factors is also determined by the environmental conditions and the amount of use the surface receives (Fleming et al. 2008; McNitt, Landschoot, & Petrukak, 2004). However, neither the influence of each surface component, nor the effect of use or environmental conditions on the surface characteristics, is fully known.

Mechanical research.

FIFA (2001) reported a number of tests for artificial surface as well as the requirements that they must attain in order for the surface to be FIFA approved. This ensures that the quality of the surface is suitable for top level matches. One such test measures the shock absorbency or cushioning of a surface. This test has traditionally been performed using the Berlin Artificial Athlete or the Stuttgart Artificial Athlete. This measures the dynamic stiffness of the surface using a falling mass (Brown, 1987) to classify the artificial sports surfaces based on the shock absorbency of a surface compared with concrete (Dixon & Stiles, 2003; Kolitzus, 1984). However, a criticism of these techniques is that the results can not be easily compared with natural surfaces, which are generally measured using a Clegg hammer. However, Fleming and Young, (2007) reported a strong correlation between the measurement of surface hardness determined with a Clegg hammer and the force reduction measured with the Berlin Artificial Athlete on a third generation artificial turf (r^2 = 0.7). This indicates that the Clegg hammer may also be a suitable method for use when assessing artificial turf.

Mechanical tests have been used by researchers to examine the effect of changes to different aspects of the third generation artificial playing surfaces. In particular, the role of the shock pad and material composition on surface behaviour has received noteworthy investigation. As indicated earlier, Martínez Durá, Gaméz et al., (2004) reported that soccer players perceive third generation artificial turf surfaces to have worse cushioning properties than natural turf surfaces. Martínez, Durá, Gaméz et al., (2004) supported these perceptions with data from mechanical cushioning tests.

The mechanical cushioning provided by artificial surfaces may change when different forms and constructions of the underlying shock pad are used. In comparing three types of shock pad to concrete, McNitt et al. (2004) found that the mechanical cushioning provided by a shock pad was influential in the reduction of forces. McNitt et al. (2004) also simultaneously measured the influence of the different shock pads whilst the surface was 'wet' and 'dry' and when varying quantities of rubber and sand were placed upon the carpet. The effect of the differences were assessed by two different mechanical tests and the authors found that depending on the type of test being performed and the ratio of sand to rubber, shock pad cushioning was influenced by whether it was wet or dry. McNitt et al. (2004) found the when 100% sand was used, the cushioning provided by three shock pads were distinctly different with the 13 mm Regupol pad providing a lower mechanical hardness compared to a 19mm E-layer pad. McNitt et al. (2004) also found that when the infill was 80% sand and 20% rubber, there was no difference in Gmax (which is the ratio of maximum negative acceleration on impact in units of gravities to the acceleration due to gravity) between three different shock pads under dry conditions and under wet conditions if a CIST equipped with a 2.25 kg missile and a drop height of 455 mm was used. However, when using the F355 method, which involved a 9.1 kg missile and a drop height of 610 mm, a 19mm E-layer pad had lower G-max values than a 13 and 19mm pad (Regupol) when under wet conditions. Therefore, the role of the shock pad changed depending on the condition of the rubber, surface moisture, and the test performed. However, at the same time, the effect of the rubber did not influence the ability of the shock pads to reduce force relative to a no shock pad condition.

A significant role has also been attributed to the infill material placed upon a new carpet in reducing the forces at impact (Fleming et al., 2008). McNitt et al. (2004) compared the effect that changes in the infill ratio would have on the forces, independent of the shock pad. They reported that when 100% rubber was used the surface was less cushioned than with the 50% sand and 50% rubber composition. Likewise, the use of 100% sand was harder than 50% sand and 50% rubber under wet and dry environmental conditions. Further still, the 80:20 sand to rubber ratio was harder than the 50% sand and 50% rubber. Therefore, the results of the mechanical tests suggest that the environmental conditions, type, material and thickness of the shock pad, as well as the proportion of the rubber crumb and sand, are important factors that can influence mechanical cushioning. These factors may therefore influence the cushioning provided to the soccer player.

The role of the shock pad has been shown to change depending on the condition of the rubber over time. Artificial surfaces can become damaged over long periods of use (Sifers & Beard, 1994). Because of this wear, the mechanical property of the infill changes as the infill material is compacted or displaced (Alcantara, Rosa, Gamez, Martinez, Comin, Such, Vera & Prat, 2006). Wear by use (number of cycles) has shown to significantly reduce the values of all parameters that define deformation and force reduction, but not vertical ball bounce (Alcantara et al., 2006). Once the mechanical properties and quantity of the infill change, the impact attenuation provided by the shock pad has been shown to become more important (Fleming et al., 2008).

Another aspect of the interaction between the shoe sole and the surface is the ability to change direction when running at speed. Traction values of between 25-50 N.M. (Traction coefficient 1.2-1.8) for good turf and between 35-45 N.M. for “ideal” natural surfaces should be achieved. Therefore, similar values should be achieved on the third generation artificial turf. To measure this, a torque wrench is used, which measures the amount of torque necessary to start motion of a studded sole (FIFA, 2001). However, there is no recognised FIFA test for measuring linear traction, although methods used on natural turf may be used on third generation turf surfaces.

Kirby and Spells (2006) analysed the effect of surface consistency by comparing ball rebound resilience and rotational resistance at different locations across the playing surface. It was shown that compared to natural turf, the artificial surface was more consistent, as the variation in the measured parameters was greater at different pitch locations on natural soccer pitches than with the artificial turf. However, Kirby and Spells (2006) found that during the late stages of the season the ball bounce and rotational resistance were outside the FIFA specifications at that time. Assuming the initial values were within the specification, this deterioration in properties had occurred in just over a six month period. This may suggest that surface damage could be more rapid than previously thought (Fleming, et al. 2008). Therefore, more frequent testing is needed to ensure surfaces meet the recommended standard.

Biomechanical research.

Although there has been some mechanical study of third generation artificial surfaces, there is very little biomechanical evidence to support the results. Meijer, Dethermer, Savelberg & Willems (2006) compared the biomechanical effect of the changes made to the surface construction and found that no differences were observed with varying elastometric infill for any of the measured parameters including peak impact force and leg stiffness. This suggested that the type of infill material (grain size and shape) does not influence the human loading, but still leaves the understanding behind infill quality and shock pad type on the performer unknown.

The responses of the soccer player to third generation artificial surfaces have also been compared to natural turf. Ford et al. (2006) measured pressure data during a cutting movement on a third generation artificial turf and natural turf surface conditions. Ford et al. (2006) found that higher peak pressures were experienced at the third and fourth metatarsals on the third generation artificial turf surfaces. Martínez Durá, Gaméz, et al., (2004) also found that compared to natural turf pitches, running on artificial soccer surfaces resulted in lower pressure across the fifth metatarsal head. However, no difference was experienced for the measurement of peak impact force.

The use of mechanical tests to compare different constructions of third generation artificial turf surfaces provides a quick assessment of the playing surface but does not account for the complex participant-surface interaction. This interaction is little understood as there are so few biomechanical studies available which compare different constructions of the third generation artificial turf surfaces. As such, much greater emphasis is needed towards understanding the movement characteristic of soccer players on these surfaces. Likewise, the surface comparisons with natural turf provides an insight into the differences between the constructions, yet little is known regarding the effect that different movements, footwear interactions or environmental conditions can have on the response of the participant to the surface. This can influence the findings on the results and can change the conclusions and recommendations regarding use of different surfaces.

2.6 Kinematic adaptation to changes in playing surface and footwear cushioning

The role of kinematic adaptations has been discussed earlier in this chapter regarding its place in the attenuation of impact forces. These same adaptations occur when participants run on different surfaces and in different footwear conditions, and can often be used to explain conflicting and confusing evidence between biomechanical and mechanical studies. McNitt-Gray, Yokoi & Millward (1993) found that peak impact force did not differ significantly when landing on gymnastic mats which had distinctly different cushioning. This was speculated to relate to adaptive strategies during landing to counteract and minimise surface effects. Similar changes are found during running and these adaptive strategies result from the ability of the participant to detect environmental change (Derrick, 2000). The body has a multitude of physiological and mechanical receptors that provide information to the brain. Derrick (2000) suggested that a normal, healthy runner will integrate this information and alter behaviour so that performance and safety are optimised.

Many of the measured biomechanical parameters that are associated with injury are dependent on the combined stiffness of the runner's geometry and surface structure. Adjustments are typically made via joint stiffness changes which allow humans to run in a similar manner on surfaces with different cushioning (Ferris et al., 1998). Further still, a playing surface is rarely uniform in its cushioning and Ferris et al. (1999) demonstrated that runners rapidly adjust leg stiffness during their first step on a new surface. By quickly adjusting leg stiffness, runners make a very smooth transition between surfaces. It has been found that in spite of a 25-fold change in surface compression, the path of the COM is almost the same before and after the transition of surfaces with no discontinuity at the transition (Ferris et al., 1999). In providing this adaptation, the biomechanical values are changed, possibly to help lower the risk of injury and optimise performance. The apparent changes in joint stiffness of the leg are assumed to occur in response to the conditions at impact and are associated with increased muscular co-activation (Gerritsen et al., 1995) or passive mechanical changes (Wright et al., 1998). Gerritsen et al. (1995) used computerised modelling simulations and found that impact forces were not sensitive to changes in muscular co-activation for realistic initial joint moments. However, the authors suggested that the results were in contrast to their hypothesis due to the limited range in which the muscle stimulation levels could be varied in the model. Peak impact forces could be substantially altered by changing muscular activations in such a way that joint moments at touchdown were not restricted to the experimental values.

As well as adjusting the leg stiffness through changes to the activation of the muscles, leg stiffness is also changed through differences in leg orientation. Changes in the perception of the surface has enabled runners to make subconscious changes to their foot position in order to lower the impact experienced (Wright et al., 1998). Wright et al. (1998) found that passive mechanical changes were evident when running on different levels of cushioning. These changes are dependent on the cushioning conditions at impact. Wright et al. (1998) observed that in hard footwear, knee flexion was increased compared to a soft shoe, but impact forces were the same. This trend is also typical for changing surfaces conditions (Gerritsen et al., 1995; Ferris et al., 1998; Ferris, et al., 1999; Stiles & Dixon, 2006). Gerritsen et al. (1995) found that in their model simulation, for every 1˚ change in initial knee angle there was a 65 N reduction in impact force. This was suspected to be because the changing knee flexion angle at contact altered the effective mass (the mass occurring vertically at the point of contact). Also, a leg with joints that are slightly flexed at ground contact will be in a better position to use eccentric muscle contractions to absorb impact energy than a leg with joints that are extended (Derrick, 2000). A lower extremity that is either too stiff or too compliant can produce unpleasant running experiences. When the leg is too stiff excessive shock will be transmitted to structures in the spine and head, and can result in excessive wear on the joint surfaces (Derrick, 2000). On the other hand, a leg that is too compliant will be unable to stop the downward progression of the body during the brief period of time that the foot is on the ground. In an extreme case this will result in a fall, but excessive knee flexion can also result in greater loads on the muscles and greater rates of oxygen consumption (Derrick, 2000). An increased knee flexion may therefore give the runner a larger margin for dealing with kinematic errors. However, the benefit is likely to have an associated metabolic cost that may impact performance (Derrick, 2004). Lafortune, Hennig, & Lake (1996) measured both VGRF and peak force transient, and found that peak impact forces were reduced with an increased knee flexion but the magnitude of the force transient remained the same. This suggests that although the peak VGRF is reduced, the forces that travel within the lower extremity are not and may therefore not be important factor when trying to lower the risk of injury.

Other adjustments have been shown to take place on less cushioned surfaces. These include increased plantar flexion, reduced heel impact velocity, and reduced initial foot sole angle relative to the horizontal (Bobbert et al., 1992; Denoth, 1986; De Wit & De Clercq, 1997; Dixon, Collop, & Batt, 2005; Gerritsen et al., 1995). Similar changes by the body have also been reported in an attempt to lower the impact loading and peak pressures (Dixon et al., 2005). Dixon et al., (2003) observed a reduction in peak ankle dorsi-flexion immediately prior to contact when a cushioning insole was mechanically degraded. Dixon et al. (2005) found a similar response to harder surfaces, with the mechanism related to the subject adopting a flatter foot at contact and local pressures can be reduced. However, such adaptations are not universally effective in reducing shock to the lower extremity (McNitt-Gray et al., 1993). For example, Dixon et al. (2005) found the use of additional knee flexion in some participants, but not all. This was shown to highlight the ability for some participants to adopt compensatory adjustments to lower peak impact loading, whereas some were unable to make these adjustments, resulting in higher impact loading on the less cushioned surfaces.

A change in magnitude to the rearfoot eversion has also been shown to reduce the impact forces and is related to the surface and footwear hardness (Stacoff et al., 1988). It has been speculated that in those studies where the material composition of the footwear had no influence on the force, the response was related to the material deformation. When the footwear mid-sole were hard or stiff, very little deformation was provided at the moment of impact and the position of the impact was towards the lateral border of the heel of the shoe (Inkaar 1994; Nigg & Bahlsen, 1988; Stacoff et al., 1988). In contrast, if the same force is exerted to the lateral heel of a soft mid-sole, the shoe becomes relatively easy to compress (Nigg & Bahlsen, 1988; Stacoff et al., 1988). The point of application of the GRF then moves to a more medial position, shortening the length of the lever or moment arm with respect to the subtalar joint axis (Stacoff et al., 1988; Nigg & Bahlsen, 1988). This more medially positioned force decreases the leverage and reduces the initial pronation, and thus increases the GRF (Kaelin et al., 1985). The speed of the initial pronation also influences the forces experienced by the athlete. While running in soft shoes, eversion velocity has been shown to be less than that of hard shoes and consequently causes the foot to ineffectively pronate. On the other hand, with the increased pronation in harder shoes, the lever arm and the distance for deceleration is increased, lowering both the impact magnitude and loading experienced. The added distance also increases the everting foot moment of rotation and the leverage along the subtalar joint axis (De Wit et al., 1995; Nigg & Bahlsen, 1988;

Nigg et al., 1987), allowing the foot to enter a more pronounced and faster initial foot pronation (De Wit et al. 1995; Nigg & Bahlsen, 1988; Stacoff et al., 1988). This change in the force vector location and pronation acceleration may therefore serve as the mechanism by which harder interfaces are able to reduce impact forces (De Wit et al., 1995; De Wit, De Clercq, & Aerts, 2000; Hreljac & Marshall, 1999; Stacoff et al., 1988).

Due to these kinematic observations, the lack of agreement between mechanical and biomechanical tests may be explained. However, the contribution of each of these factors is individual and some participants will not experience kinematic change. Therefore, the individual running strategy will influence the magnitude of GRF when running on differently cushioned surfaces and this can cause the GRF to be greater in some participants and lower in others.

Another reason for the lack of significant differences could be related to the measurement of GRF. As mentioned earlier, GRF is the measurement of acceleration to the body's COM. The initial conditions at touch down influence the way the segments are decelerated, dictating the characteristics of GRF at impact (Bobbert et al., 1992). Bobbert, Schamhardt & Nigg, (1991) showed that while the first peak is passive and has its origin in the high frequency accelerations of the lower extremity segment/support leg, the absolute force value of the first peak in vertical force depends not only on the passive contribution of the support leg, but also on the contributions from the head, arms and trunk and swing leg. These are summed as 'the rest of the body' and their contribution is far bigger due to a larger mass rather than a high acceleration. Therefore, Bobbert et al. (1992) illustrated that the VGRF describes the forces acting to control the vertical acceleration of the total body's COM acting at the foot/ground interface which may be influenced by other factors apart from the ground. Therefore, it does not solely reflect the changes in accelerations of the lower extremity as these accelerations and its specific mass only account for a part of the total acceleration to the body's COM. As such, changes to the surface or footwear may be ineffective in significantly altering peak impact forces

As well as the effect of the contribution of the 'rest of the body' to the overall magnitude of the GRF, VGRF is distributed spatially, temporally and in the frequency domain (Hamill, 1996). The true magnitude of the impact force at the heel may be masked by the superimposition of low frequency active components from muscular action and forces acting on other parts of the foot. This superimposition has been found in leg shock measurements. Shorten & Winslow (1993) demonstrated that the magnitude of the passive impact is not uniquely determined by the high frequency heel impact component. Both low and high frequency motions of the body's COM and loads applied to other regions of the foot contribute to the peak (Hamill, 1996). Because of this, the measurement of VGRF reflects the forces applied to the COM and does not necessarily describe the passive motion of the foot at the foot-ground interface. The accelerations of the support leg influence the magnitude of the impact peak, so decomposition of the vertical time-history at impact can show the passive phase of the graph (Hamill, 1996). This may be a more relevant measure by which the relationship between high magnitudes and injury may be established and also to establish differences between cushioning, although is more difficult and time consuming.

In summary, kinematic adaptations can be used to change the alignment of the lower limb so that the lower the peak impact, loading rates and pressure experienced. Likewise, the lack of significant difference between differently cushioned footwear and surfaces when measuring GRF may relate to the lack of sensitivity of GRF measurements have to changes in footwear and/or surface conditions.

3. Study One:
The Biomechanical Comparison of Footwear-Surface Combinations in Soccer and the Effect of Seasonal Variations

3.1. Introduction

Injury rates fluctuate throughout the year in soccer, and are found to be greatest during the periods termed preseason and early season (Woods et al., 2002; Woods et al., 2003). At this time, natural surface conditions are reported as being harder and drier than at other periods of the year (Woods et al., 2002). This is consistent with claims that the quality of natural playing surfaces is a risk factor for injury in soccer (Hawkins et al., 2001; Junge et al., 2004; Yde & Nielsen, 1990).

One way in which the risk of injury may be changed is by replacing the use of natural turf with an artificial surface. Third generation artificial turf has been developed to be a soccer specific artificial surface, which is more consistent over time in its characteristics than natural turf. This may therefore be an appropriate alternative to natural turf in summer. Conversely, the characteristics of this artificial surface may be unfavourable during other periods of the year when conditions on natural turf are perceived as optimal for soccer (with the exception of waterlogged or frozen pitches that occur sporadically during this time). Characteristics of third generation surfaces have been compared to natural turf using biomechanical techniques (Ford et al., 2006; Martínez, Durá, Gaméz et al., 2004), but these studies have failed to take into consideration the effect that seasonal change to the natural turf may have on conclusions drawn from the comparison.

Soccer players also require consideration of the type of soccer boots worn as these have also been linked to injury (Woods et al., 2002). Modern soccer boots are designed with different plantar sole constructions, including screw-in and moulded studs, and soccer trainers. Biomechanical research has evaluated some different stud configurations (Dixon et al., 2008l; Queen et al., 2008; Coyles & Lake, 1999). Since there are many different brands and alternative stud configurations, further research is needed. Further still, the construction and patterns of the plantar studs have been developed to suit different surface conditions (Getz & Brannan, 2007). Therefore, the choice of footwear for a particular surface condition is a key consideration for participants as it may influence the level of injury risk (Torg & Quadenfeild, 1971). As such the surface and footwear likely interact to determine the biomechanical response of the participant.

To understand why injury rates differ at different times of the year, it is important to evaluate the biomechanics of players at different periods of the year. This has not previously been conducted in biomechanical studies of natural turf surfaces. Further still, despite third generation turf being considered consistent over time, various researchers have speculated that amongst other factors, surface moisture and the condition of the rubber may influence the mechanical response of the surface (Fleming et al., 2008; Kirkby & Spells, 2006; McNitt et al., 2004). While surface moisture did not influence mechanical surface hardness (McNitt et al., 2004), the quantity and compaction of rubber did (Fleming et al., 2008; McNitt et al., 2004). This can occur over time with an ageing surface, yet at present, no biomechanical data have been collected to see how the age of third generation surfaces may influence the behaviour of the soccer player.

Of the injuries present in soccer, non-contact, acute ankle injuries are common and occur during turning movements (Woods et al., 2003). Acute damage to the lateral ankle ligaments, particularly the ATFL, can result from excessive footwear-surface traction (Ekstrand & Nigg, 1989; Nigg & Segesser, 1988, 1992). During turning and cutting movements, increased traction generated by decreased surface moisture and increases in stud length accentuate the magnitude and loading rate of horizontal forces applied to the ankle, which can amplify the magnitude of the inversion movement. This in turn increases the loads on the lateral ligaments, peroneal muscles and connective tissues (Durá et al., 1999), contributing towards their injury. As a consequence, surface and footwear changes may change the peak inversion angle and may influence the risk of acute injury whilst performing movements such as turning.

The footwear-surface combinations may also influence the susceptibility to overuse injury, where changes to these factors may influence the cushioning during running and turning. Specifically, cushioning is suspected to be reduced as the surface dries and stud length increases. Surface and footwear cushioning has been commonly assessed using the magnitude of peak impact force as a measure of lower extremity loading. Despite this, researchers have found it difficult to distinguish between the cushioning of mechanically differently surfaces and footwear conditions, using this measure during a running task. On the other hand, differences have been observed during more dynamic movements (Stiles & Dixon, 2006), thus advocating the use of alternative movements such as turning when assessing surfaces. Researchers have also measured peak impact loading rate, peak pressures and peak pressure loading rate which are also linked to the occurrence of overuse injury. In contrast to peak impact force, these measurements have been used to distinguish differences between surfaces and footwear of varying mechanical hardness (Dixon et al., 2008; Shorten, 2002; Queen et al., 2008). This may indicate that these measurements are more sensitive to changes in surface and footwear cushioning.

Kinematic adaptations have also been used to explain the consistency of peak impact force between surfaces of different mechanical cushioning. Greater rearfoot, ankle and knee angles have been shown in participants in response to harder playing surface conditions. These changes have been suggested to be compensations to allow maintenance of similar peak force when running across surfaces of different cushioning (Dixon et al., 2005; Ferris et al., 1998; Whiting & Zernicke, 1998). Kinematic changes can also influence forces during lateral movements. Increased rearfoot inversion can significantly reduce peak impact force (Dayakidis & Boudolos, 2006), although this can magnify the loading of the lateral ligaments causing increased pain during these movements (Luethi et al. 1986). Likewise, increased plantar flexion can help attenuate impact forces during turning (Brizuela et al., 1997), although this increase also allows the level of inversion experienced to be exacerbated (Fujii et al., 2005) and increases the risk of ATFL injury. Overuse injury to the Achilles tendon has been associated with the magnitude of dorsi-flexion, thus it could be assumed that on hard playing surfaces, if kinematic changes increase the magnitude of dorsi-flexion, this would likely predispose the soccer player to increased injury risk.

3.1.1. Aims and Hypotheses

There are three main research questions that will be addressed in this chapter. The first question addresses the extent to which the environmental conditions influence the biomechanical comparison of soccer players on natural turf and third generation surfaces. To assess this question a comparison is made of a third generation artificial surface to natural turf at two periods of time. It is hypothesised that the lower extremity loading will be significantly greater when running and turning on the artificial surface compared to the natural turf when the environmental conditions are wet and cold. This is because the sodden soil will act as a more effective shock absorber than the third generation turf, perceived by some as being harder than natural turf (Martínez, Durá, Gaméz et al., 2004). To assess this lower extremity loading it is hypothesised that peak impact force (particularly during turning), peak loading rate, peak pressure (medial and lateral heel and first and first metatarsals) and peak pressure loading rate (medial and lateral heel and first and first metatarsals) will be greater (during running and turning) on the artificial turf. Likewise, during turning, peak rearfoot inversion will be greater on the artificial turf surface because the traction provided by the artificial turf will be greater as it provides a more efficient drainage of moisture lessening footwear slide. However, these biomechanical measurements will be reversed on the natural turf compared to the third generation artificial turf when the environment is warm and dry. This is because the environmental conditions will encourage the natural surface to dry and become harder resulting in decreased cushioning and increased traction compared to the third generation turf, which is less influenced by the environmental conditions.

The second research question will address how the biomechanical characteristics of soccer players change in different soccer boots and how these differences are influenced by performing on different surfaces at different times of the year. It is hypothesised that the lower extremity loading, specifically peak impact force (particularly during turning), peak loading rate, peak pressure (medial and lateral heel and first and first metatarsals) and peak pressure loading rate (medial and lateral heel and first and first metatarsals) will be reduced in the moulded rubber studs compared to the metal screw-in studs and in a soccer trainer compared to the moulded and screw-in. Likewise, peak rearfoot inversion will be significantly greater when participants wear the longer screw-in studs compared to a moulded boot and soccer trainer. It is further hypothesised that an interaction will occur between the soccer boot and the surface which will significantly influence the biomechanical response of the soccer player. In particular, whilst wearing the screw-in studs on the artificial surface, significantly greater lower extremity loading will occur when running and turning compared to the other footwear-surface combinations when the environment is wet and cold. Likewise the combination of third generation artificial surface and screw-in studs will cause greater rearfoot inversion and plantar flexion when turning at this time. On the other hand, when the environment is dry and warm, these trends will be true on the natural surface in the studded boots, compared to the other footwear-surface combinations.

The third research question in this study will address the extent to which biomechanical changes occur on natural and third generation artificial turf surfaces between two times of the year when the moisture and temperature are different. It is hypothesised that there will be significantly greater peak impact force (particularly during turning), peak loading rate, peak pressure (medial and lateral heel and first and first metatarsals) and peak pressure loading rate (medial and lateral heel and first and first metatarsals) when running and turning on the natural turf when the environment is warm and dry compared to a wet and cold environment. However, it is hypothesised that the third generation artificial turf surface will stay consistent between tests, resulting in no significant differences.

3.2. Methods

3.2.1. Participants

Seven male participants (age 21.7 [S.D. 2.2] yrs, weight 74.0 [S.D. 6.9] kg [March] 74.6 [S.D. 6.9] kg (May), footwear size 10 -11) were recruited for this research investigation. Participants were active soccer players at a local league level and were free from any lower limb injury during the three months prior to the start of the testing. The participants were made aware of this exclusion criterion and were told that they could drop out of the research project at any time, particularly if they were to become injured. Following this, all participants agreed to take part in the research investigation, understood the nature of the testing and the time commitment that was needed. To confirm this, each participant signed an informed consent form which along with the investigation was approved by the ethics committee at the School of Sport and Health Sciences, University of Exeter.

In order to collect kinematic data, participants wore reflective joint markers, carefully positioned upon the key points of interest (hip, knee, ankle of boot, and fifth metatarsal). These were used to indicate the thigh (hip and knee), shank (knee and ankle of boot) and foot (lateral maleolus and fifth metatarsal) segments (Figure 3.1). Additionally, two markers were added to define the line of the calcaneous, and two more to define the line of the Achilles tendon/tibia, allowing the monitoring of rearfoot angle (Figure 3.2).

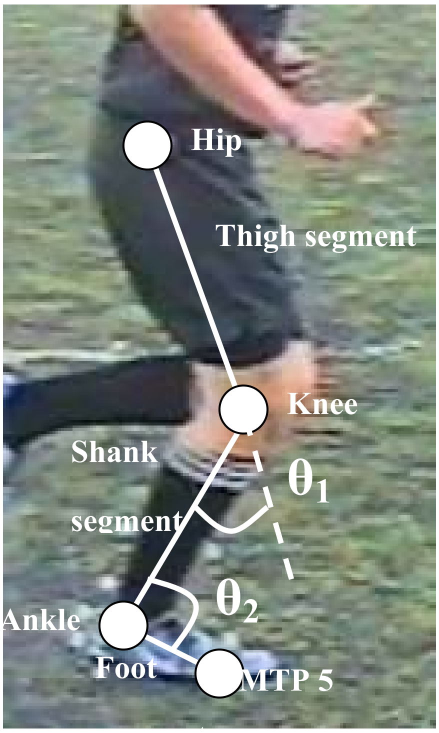


Figure 3.1. A view of the marker positions in the sagittal plane

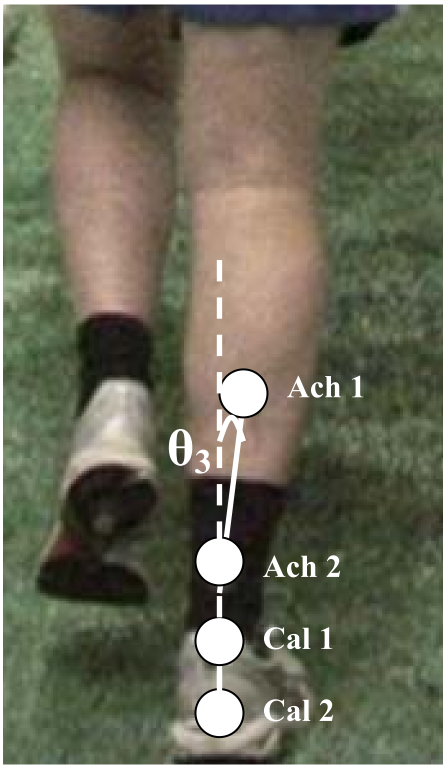


Figure 3.2. A posterior view of the marker positions in the frontal plane

3.2.2. Independent variables.

The experimental conditions included two playing surfaces and three types of footwear. The surfaces included a council owned third generation artificial soccer pitch and a natural turf playing surface. The three footwear models used were similar in construction and material of the upper, but different in the plantar sole design and material composition and stud configuration. These included boots with 6 screw-in studs (Nike Total 90, Figure 3.3), 15 rubber moulded studs (Nike Total 90, Figure 3.4) and a pair of soccer trainers (Nike, Figure 3.5).

Figure 3.3. Example of screw-in soccer boot

Figure 3.4. Example of moulded soccer boot

Figure 3.5. Example of soccer trainer

The same studded footwear and surfaces were tested on two occasions (March and May). These times corresponded to periods of the year when wet and cold and warm and dry climate conditions are experienced in the south west region of the UK (Met Office, 2006). The soccer trainer condition was tested on the natural surface and third generation artificial surface in May only. Both the natural and artificial playing surfaces were not treated or prepared by the researcher. This meant that the moisture or contours of the surface were not necessarily uniform, replicating potential conditions found by soccer players.

Each participant was assessed on the same natural and artificial turf surface but ran across a different area of the pitch. This was important because the material of both the natural and artificial turf may become more compact as the number of trials progressed causing a hardening of the surface, which along with the composition of the surface, may influence the forces experienced by the participants (Stiles et al., 2008; Fleming et al., 2008). To further limit the effect of surface compaction, each footwear condition was tested in a random order for each participant. This ensured that any change in surface conditions that did occur within one test condition did not have a systematic effect on the biomechanical measurements taken with each footwear type.

Prior to any biomechanical data collection, mechanical testing was performed on each surface every time a participant was tested by using a Standard 0.5kg Clegg hammer (Model 500GT) (Dr Baden Clegg Pty Ltd, Australia). This allowed the mechanical quantification of the surface cushioning and any change that occurred over time. The Clegg hammer is a device that has a 0.5 kg weight attached to an accelerometer which is placed into a tube. The weight is then dropped from a height of 30 cm five times, the fifth being recorded as the measurement of surface cushioning (Dr Baden Clegg Pty Ltd, Australia). This cushioning is reported as peak gravities or G (multiple gravities) (see Chapter 2: Figure 2.18).

3.2.3. Procedure

As the research focuses upon movement and pressure characteristics during running and turning, participants' performed eight trials for each footwear-surface condition and kinematic, force and pressure data were collected simultaneously. This number of trials ensured reliable pressure insole data during running and turning (Appendix B & C). Each participant was asked to perform habituation trials. This involved the participants performing the running and turning movements that replicated a test trial, until they were familiar and comfortable with the test protocol. This protocol was as follows. Each trial was performed within a designated area, nine meters in length. This area was determined by a marked line that represented a start line (A) and another to represent an end line (D) (Figure 3.6). Within this space two other lines were defined. The first was a line marked three meters from the start (B) and the second a line marked at six meters from the start (C).

Participants started at (A) and ran the length of the test area (up to line D). At approximately mid way (between lines B and C), a square of $1m^2$ in size was marked. Whilst running between lines (A) and (D), participants were asked to place their right foot into this marked box without adjusting their natural running gait (3.0 m/s ± 5%). The speed of this component of the trial was standardised using three photosensitive timing gates one of which was located at the start line and two others placed 1.5 metres either side of the marked box at location B and C (Figure 3.6). The timing of the trial speed started when the participant broke the infrared beam emitted by the first gate. As the participant passed through the subsequent gates the beam from these gates were broken, enabling the time to be recorded for each distance. At the end line the participant placed their right foot on the marked line, then twisted their hip and pushed off towards the start line. The participants were then told to continue running at the same speed until they reached the 3 metre line (B). Here participants were asked to turn as they had done at the end line and run towards the boxed area. Once approaching this area the participant prepared to make their final turn, which involved them placing their right foot within the marked area and turning as previously directed. This was at a self selected speed so that to ensure the athletes felt comfortable with the procedure without putting them at increased risk of injury. They then completed the trial when they crossed the start line (A). The average speed of the remainder of the trial was standardised by monitoring the time taken from the start of the trial to the end of the trial. This was specific to the individual but was consistent for each trial. By calculating the total length of time taken, the assumption was made that the remaining components of the trial including the turning movements was at a consistent speed. Any trial where either the straight run or total time was not at the required speed or where the movement pattern was not as directed was subsequently repeated.

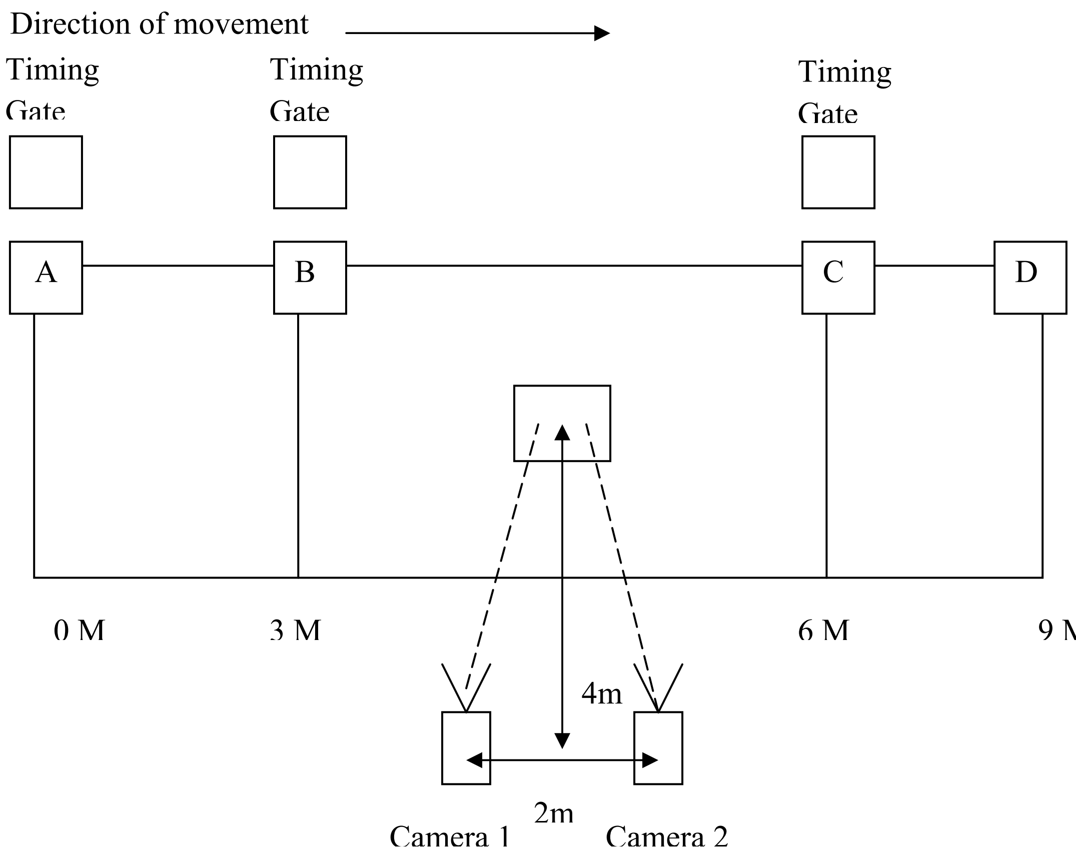


Figure 3.6. Schematic of the experimental test area, where M = distance in metres and lines A and D represent the start and end lines of the test area respectively. Within this space, line B was used to indicate changes in movement direction. Lines B and C were also the location of the timing gates.

3.2.4. Data collection

3.2.4.1. Kinetic.

To collect kinetic data, a pair of Footscan pressure insoles (RSscan International, Belgium, 500 Hz) were inserted into the test footwear (Figure 3.7). The insoles are designed to be flexible and fit flat into the footwear without compromising the normal gait pattern of the participant. These insoles extended from the lateral edge and when worn in the soccer boot, rested up against the lateral side of the leg. The insole was fastened with two Velcro ties to avoid interference with the gait of the participant. This material allows the electric impulse emitted by the piezo-electric crystals to be transferred from the insole to a data logger worn by the participant around the waist (Figure 3.8).

Figure 3.7. An example of a pressure insole worn by the participants

Inserted into the data logger was a capture card that collected seven seconds of pressure and force data. This was activated by the investigator via a remote control (Figure 3.8) at the moment the participant started the trial. Once the trial was complete the capture card was transferred to an external device connected to a lap top. Transferred data were then saved from the card to the computer for future analysis. The capture card was then cleared so that the next trial could be stored.

Figure 3.8. An example of the data logger and remote control activator

3.2.4.2. Kinematic.

The experimental set-up for collecting kinematic data was as follows. Two 25 Hz VHS cameras (Panasonic, SVHS, AG-DP200) were set up with a 2:1 base ratio. This meant that the cameras were positioned two meters apart and were set parallel to the test area (Figure 3.6). The mid-point between the two cameras was used as a reference point to measure the location of the test area. From here, the test area was four meters square to this line. As such the ratio of the distance between the cameras is half of the distance between the camera mid-point and the test area. The cameras were positioned on tripods, set at a height of 1.5 meters, and were angled parallel to the ground. Each participant then stood in the square and the cameras were zoomed and focused so that each key point of interest could be clearly seen by each camera. Finally, a calibration object with 12 markers was placed to cover the test and surrounding area, at a height of 1.2 m so that each marker could be clearly seen by each camera (Figure 3.9).

Figure 3.9. An example of the calibration frame used to calculate the DLT parameters

Once the object was filmed, the cameras were not moved. To ensure this, a spirit level was used frequently throughout the testing to check that no changes had occurred. Participants then stood in the test area to calculate the segmental angles for a standing position.

After filming the participant's trials, the footage was transferred to a computer where, each trial was matched from each camera and were synchronised by the moment of heel contact with the ground. The movement was then digitised manually for both camera views as was the calibration frame using Hu-m-an movement analysis software (HMA Technology Inc, Ontario, Canada). Three-dimensional co-ordinates were calculated for each marker using a direct linear transformation (DLT).

3.2.5. Data analysis

Once the three-dimensional co-ordinates were reconstructed, rearfoot movement occurring in the frontal plane was calculated by measuring the angle between the line representing the calcaneous relative to the line representing the tibia (Figure 3.2). The calculated angle time-history was smoothed using a three point moving average. A three-point moving average was used as other methods such as the quintic splines require a greater number of data points. Since the cameras only collected data at 25Hz, an insufficient number of frames were supplied to apply an alternative technique. Peak rearfoot eversion (the greatest negative value) was taken from the data collected during running, and peak inversion (the greatest positive value) during turning.

Smoothed peak knee flexion and ankle plantar flexion and dorsi-flexion were measured using the sagittal plane co-ordinates (Figure 3.1). The knee flexion was calculated by measuring the angle of the shank relative to the thigh. Ankle dorsi-flexion and plantar flexion were calculated by measuring the angle between the foot and the shank.

To quantify the plantar foot pressures, the Footscan software (version 6.345) was used and masks were set by the software at the location of the medial and lateral heel along with the first and fifth metatarsal (Figure 3.10). The peak pressures and peak pressure loading rate at the medial and lateral heel, as well as the first and fifth metatarsal, were then taken from the software and statistically analysed. Also extracted from the software were the peak forces at impact (Figure 3.11). Peak rate of loading was measured using the gradient of the slope leading to the peak vertical force. This was calculated with the use of the first central difference method (See equations 3.1 and 3.2).

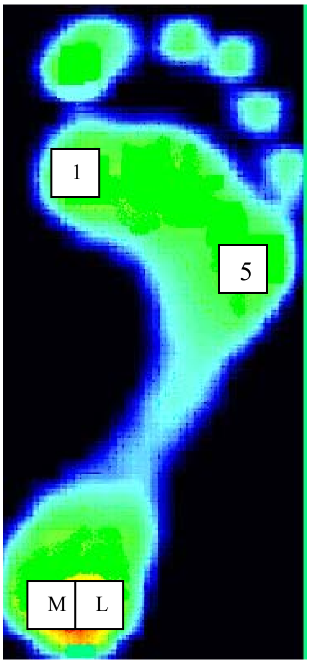


Figure 3.10. An example of pressure mask location set by the Footscan software, where M and L are the mask at medial and lateral heel respectively, and 1 and 5 are masks at the first and fifth metatarsal respectively.

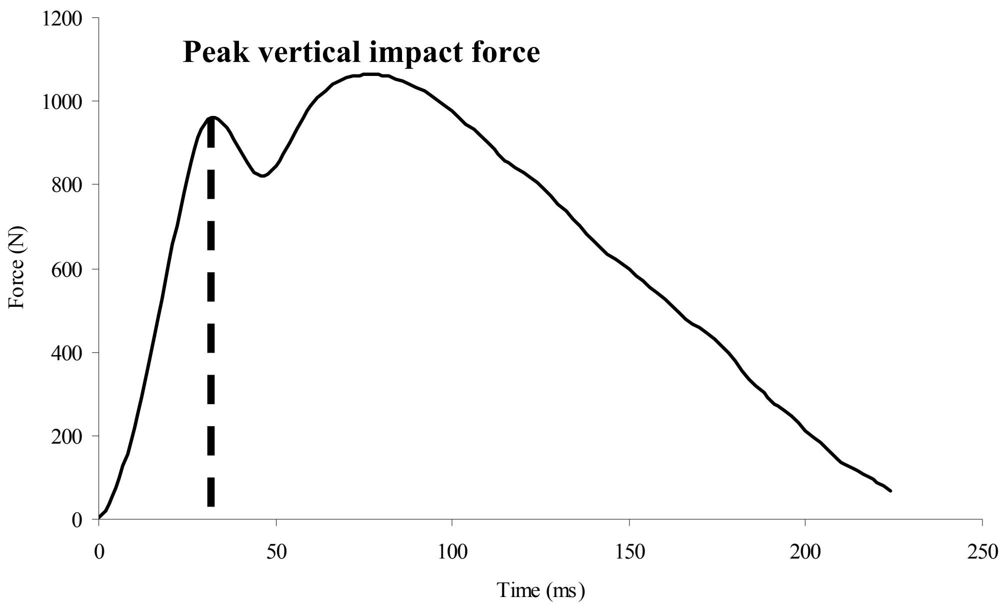


Figure 3.11. An example of vertical impact force time-history and the location of the peak vertical impact force.

$L_1 = (x_i+1 - x_i-1) / 2\Delta t$ Equation 3.1

Where:
Li = instantaneous loading
x_i = vertical force for the ith field
Δt = time interval

During the first data collection period in March, one participant dropped out of the research investigation due to injury, prior to collecting data on both surfaces. Due to this, the initial data provided by the participant was removed from the data set. Also due to technological problems, pressure data was only collected for participants 1-4 for all conditions during test two. The pressure data provided by participant 5 and 6 was used when comparing the conditions in March but were removed when comparing between test sessions. As the trainer condition was only used during May, this data were also removed from the data set during the comparison between sessions.

3.2.6. Statistical analysis.

Statistical analysis was performed using SPSS (15.0 for Windows). Using the mean result of the 8 trials collected for each condition, an Analysis Of Variance with repeated measures (ANOVA RM) highlighted any significant differences between the test variables. Two separate ANOVA RM were performed to compare the data collected during March and May. For the data collected during March, a 2X2 ANOVA RM was performed, where the independent variables measured were two playing surfaces (natural turf and third generation artificial turf) and two footwear (screw-in and moulded). In May, a 3X2 ANOVA RM was performed, where the independent variables included two surfaces (natural turf and third generation artificial turf) and three footwear conditions (screw-in, moulded and soccer trainers). Sphericity of the data was tested by analysing the Mauchly's tests of Sphericity. Sphericity was assumed if this was not significant (at alpha level >0.05). However, if this statistic was significant, a Greenhouse Geisser correction was used when comparing the main test effects.

Paired samples t-tests with bonforroni adjustments were used to highlight the location of any significant differences. To compare the surfaces between tests, paired samples t-tests were performed, one of which compared the biomechanical measurements collected on natural turf on test 1 (March) against measurements on natural turf on tests 2 (May) and the other to compare the third generation turf on these occasions. The alpha used for statistical significance was $p < 0.05$.

3.3. Results

3.3.1. Mechanical tests

The hardness of both the natural turf and the third generation artificial surface was quantified each time a participant was tested using the Clegg hammer. The mean mechanical hardness of each surface in March, indicated that the third generation artificial surface was harder than the natural turf (93.7 ± 3.2g, compared with 80.0 ± 4.0g). The mechanical tests were also repeated on both surfaces during the second testing period, where the natural playing surface was harder than the third generation artificial surface (102.0 ± 3.0g and 96.0 ± 2.7 respectively) although the difference between the surfaces was less than that observed during data collection in March. The data also showed that the natural surface was harder on the second test occasion compared to the first.

The climatic conditions were collected from the Met office records (Table 3.1). Compared to measurements collected between 1971 and 2000, the average maximum temperature for both March and May 2007 was typical. Likewise the rainfall experienced during March replicated previous years, although the rainfall data collected during May was shown to be greater than average values.

Table 3.1 *Mean maximum temperature and rainfall occurring across the south west of England during March and May 2007*

	Max temp [°C]	Max temp [°C]	Rainfall [mm]	Rainfall [mm]
Years	2007	1971-2000	2007	1971-2000
March	10.9	9.4	96.8	101.2
May	15.9	15.1	137.8	71.8

3.3.2. Biomechanical data (turning) - Test 1 (March)

3.3.2.1. Kinetic.

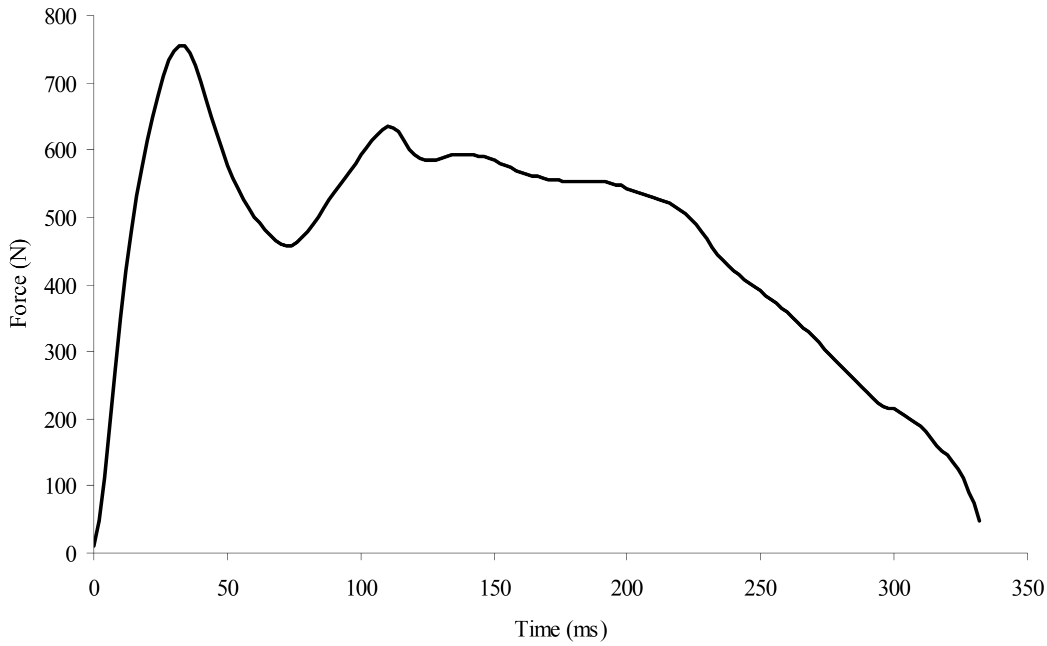


Figure 3.12. A typical force-time history graph for the vertical force during turning

Figure 3.12 shows the trace of a typical force-time history during turning. The first peak represents the peak vertical force. Table 3.2 presents the data collected whilst turning on the natural and third generation surfaces during test 1. It is shown that there were no significant differences for the measurement of peak forces and peak loading rates occurring whilst turning on the different playing surfaces. However, peak pressures at the lateral heel are significantly greater on the natural turf surface, whereas the peak pressure loading rate was larger at the medial heel. No significant difference was observable between surfaces for peak pressures at the first and fifth metatarsal (Table 3.2). The statistical analysis showed that the footwear were not significantly different (Table 3.3). There was a significant interaction indicated between the footwear and the playing surface conditions (Table 3.4) for the measurements of peak pressure and peak pressure loading rate at the first metatarsal. This was shown to be greater in the studded boot compared to the moulded on the third generation artificial turf yet no difference was observed between the footwear on the natural turf. A similar interaction was present for the peak pressure loading rate at the first metatarsal, but post hoc tests showed that these individual differences were not significant.

Table 3.2 P*eak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected whilst turning on the two surfaces (natural and third generation artificial turf (3g)) during March*

	Surface	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	609.92 (67.15)	0.54
	Natural	620.86 (91.65)	
Peak rate of loading (N/ms)	3g	23.53 (15.53)	0.11
	Natural	18.62 (5.62)	
Peak medial heel pressure (H1) (N/cm^2)	3g	20.06 (4.67)	0.64
	Natural	28.97 (11.76)	
Peak lateral heel pressure (H2) (N/cm^2)	3g	6.02 (2.44)	**0.04***
	Natural	5.44 (3.26)	
Peak first metatarsal pressure (M5) (N/cm^2)	3g	16.39 (3.05)	0.46
	Natural	26.37 (13.32)	
Peak fifth metatarsal pressure (M1) (N/cm^2)	3g	19.36 (5.90)	0.25
	Natural	23.19 (17.69)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	0.75 (0.16)	**0.03***
	Natural	1.09 (0.55)	
Peak lateral heel loading rate ($N/cm^2.ms$)	3g	0.57 (0.13)	0.27
	Natural	1.13 (1.31)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	0.64 (0.24)	0.20
	Natural	0.77 (0.39)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	0.17 (0.07)	0.55
	Natural	0.15 (0.10)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level.

Table 3.3 *Peak impact force, peak rate of loading, peak pressures and peak pressure loading rate data collected during March whilst turning in the footwear (moulded and screw-in)*

	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	Moulded	621.93 (77.47)	0.37
	Screw-in	608.85 (82.94)	
Peak rate of loading (N/ms)	Moulded	17.62 (4.77)	0.25
	Screw-in	24.53 (15.76)	
Peak medial heel pressure (N/cm^2)	Moulded	25.04 (8.93)	0.18
	Screw-in	23.99 (11.10)	
Peak lateral heel pressure (N/cm^2)	Moulded	20.25 (9.30)	0.16
	Screw-in	22.52 (12.31	
Peak first metatarsal pressure (N/cm^2)	Moulded	19.57 (7.26)	0.68
	Screw-in	22.97 (17.22)	
Peak fifth metatarsal pressure (N/cm^2)	Moulded	4.85 (2.75)	0.45
	Screw-in	6.62 (2.74)	
Peak medial heel loading rate ($N/cm^2.ms$)	Moulded	0.85 (0.36)	0.23
	Screw-in	0.10 (0.50)	
Peak lateral heel loading rate ($N/cm^2.ms$)	Moulded	0.64 (0.31)	0.20
	Screw-in	1.06 (1.31)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	Moulded	0.75 (0.42)	0.54
	Screw-in	0.66 (0.21)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	Moulded	0.15 (0.10)	0.64
	Screw-in	0.17 (0.07)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability

Table 3.4 *Interactions between the playing surface on (natural and third generation artificial turf (3g)) and footwear (moulded and screw-in) whilst turning during March for measurements of peak impact force, peak rate of loading, peak pressures and peak pressure loading rate*

	Playing Surface	Footwear	Mean and standard deviation (S.D)	P
Peak Impact Force (N)	3g	Moulded	647.09 (44.40)	0.30
	3g	Screw-in	572.75 (68.05)	
	Natural	Moulded	596.78 (98.55)	
	Natural	Screw-in	644.94 (85.88)	
Peak rate of loading (N/.ms)	3g	Moulded	17.79 (3.60)	0.41
	3g	Screw-in	29.27 (21.51)	
	Natural	Moulded	17.46 (6.08)	
	Natural	Screw-in	19.79 (5.42)	
Peak medial heel pressure (N/cm^2)	3g	Moulded	21.32 (5.65)	0.57
	3g	Screw-in	18.79 (3.49)	
	Natural	Moulded	28.76 (10.50)	
	Natural	Screw-in	29.20 (13.92)	
Peak lateral heel pressure (N/cm^2)	3g	Moulded	16.06 (2.87)	0.31
	3g	Screw-in	16.72 (3.47)	
	Natural	Moulded	24.43 (11.84)	
	Natural	Screw-in	28.31 (15.53)	
Peak first metatarsal pressure (N/cm^2)	3g	Moulded	4.11 (1.39)	**0.01***
	3g	Screw-in	7.94 (1.54)	
	Natural	Moulded	5.60 (3.66)	
	Natural	Screw-in	5.29 (3.15)	
Peak fifth metatarsal pressure (N/cm^2)	3g	Moulded	17.95 (6.50)	0.92
	3g	Screw-in	20.77 (5.44)	
	Natural	Moulded	21.21 (8.21)	
	Natural	Screw-in	25.17 (24.73)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	Moulded	0.72 (0.21)	0.35
	3g	Screw-in	0.78 (0.09)	
	Natural	Moulded	0.98 (0.46)	
	Natural	Screw-in	1.20 (0.66)	

Peak lateral heel loading rate ($N/cm^2.ms$)	3g	Moulded	0.53 (0.14)	0.36
	3g	Screw-in	0.61 (0.12)	
	Natural	Moulded	0.75 (0.13)	
	Natural	Screw-in	1.52 (1.80)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	Moulded	0.13 (0.07)	**0.01***
	3g	Screw-in	0.21 (0.05)	
	Natural	Moulded	0.17 (0.13)	
	Natural	Screw-in	0.12 (0.67)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	Moulded	0.62 (0.31)	0.17
	3g	Screw-in	0.65 (0.17)	
	Natural	Moulded	0.87 (0.50)	
	Natural	Screw-in	0.67 (0.26)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level.

3.3.2.2. Kinematic.

Two-dimensional sagittal and frontal plane kinematics were calculated from the three-dimensional reconstructed marker co-ordinates. Table 3.5 presents the potential error in the calculation of the angles due to the inability to mechanically synchronise the cameras. To account for this possible error, a sensitivity analysis was performed. The maximum difference in angle for the rearfoot, knee flexion and ankle plantar flexion and dorsi-flexion was compared between the trials. The first calculation of these angles used the moment of ground contact to match both cameras. Two other angles were calculated using the same video footage but where the first angle used the moment of ground contact, these angles used the moment of ground contact from one camera and matched it to the frame before and after ground contact from the other camera. This was calculated for 8 running and 8 turning trials. The peak angles were then calculated when synchronisation was made between two cameras at heel contact (0), when one camera is at heel contact and the other a frame before heel contact (-1) and when one camera is at heel contact and the other is a frame after heel contact (+1). These differences are presented in Table 3.5 and are non-significant, indicating that the lack of mechanical synchronisation does not significantly influence the peak angular data obtained.

Table 3.5 *Peak knee flexion, plantar flexion, dorsi-flexion, inversion (during turning) and eversion (during running) angles calculated with three methods of camera synchronisation. Method one matches two cameras at heel contact (0), method two matches one camera at heel contact to the second camera at a frame before heel contact (-1) and the third method matches one camera at heel contact to the second camera a frame after heel contact (+1)*

		Knee flexion (degrees)	Ankle plantar flexion (degrees)	Ankle dorsi-flexion (degrees)	Maximum rearfoot movement (degrees)
					Eversion
Running	-1	20.7 (9.2)	127.3 (15.0)	108.6 (11.3)	16.6 (6.9)
	0	18.8 (11.7)	126.8 (12.2)	108.8 (12.0)	16.0 (5.5)
	+1	19.4 (9.6)	124.0 (9.10	109.7 (10.9)	16.0 (5.7)
P		0.30	0.44	0.34	0.81
					Inversion
Turning	-1	62.2 (28.6)	97.3 (14.0)	83.8 (14.5)	-22.1 (16.1)
	0	63.6 (30.1	97.5 (15.6)	85.7 (12.3)	-22.4 (13.8)
	+1	59.8 (27.7)	98.5 (19.2)	88.5 (12.7)	-20.7 (14.0)
P		0.10	0.90	0.44	0.73

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Figures 3.13, 3.14 and 3.15 are typical time-histories of knee, ankle and rearfoot angles respectively. Participants were shown to exhibit an almost straight knee at impact and then the angle became greater as the knee flexed. Maximum knee flexion was represented by the peak in the time history during mid-stance. The knee angle was then reduced as the participant pushed out of the turn. The shape of the typical time history of the ankle joint angle showed that the participants performed an initial plantar flexion of the ankle where it peaked and then moved into a dorsi-flexed position towards the end of ground contact. At this point, the ankle plantar flexes as the participant pushes out of the turn. Finally, the typical time history of the rearfoot showed that the participants contacted the ground and immediately inverted the rearfoot. At a peak inversion position, the participant everted the rearfoot, until maximum eversion occurred towards the end of the turning action. At this point the participant performed a re-inversion movement during the propulsive phase of the action.

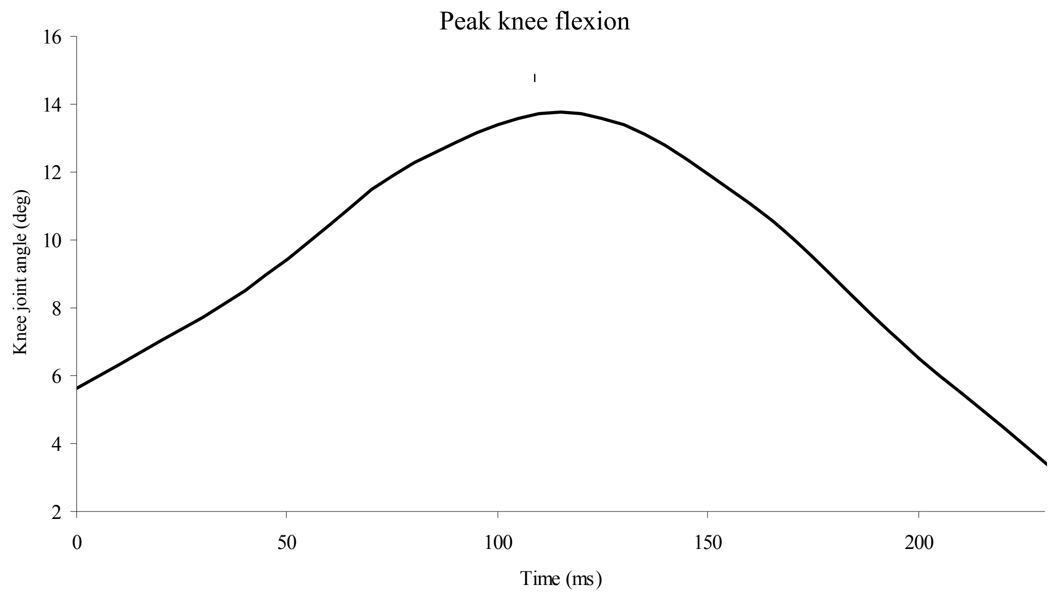


Figure 3.13. An example angle-time history showing of peak knee flexion-extension when turning

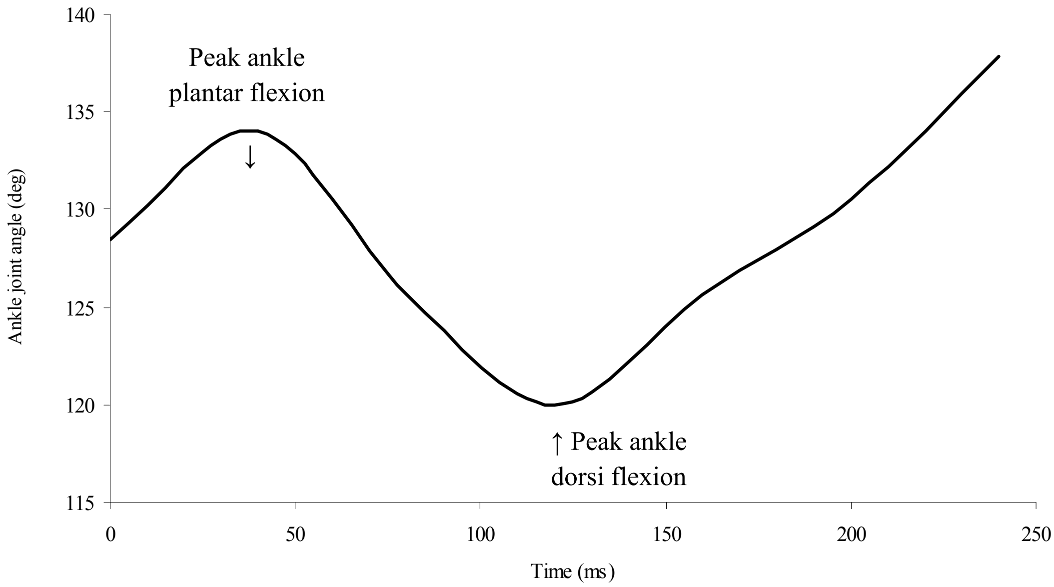


Figure 3.14. An example angle-time history showing of peak ankle dorsi-flexion and plantar flexion when turning

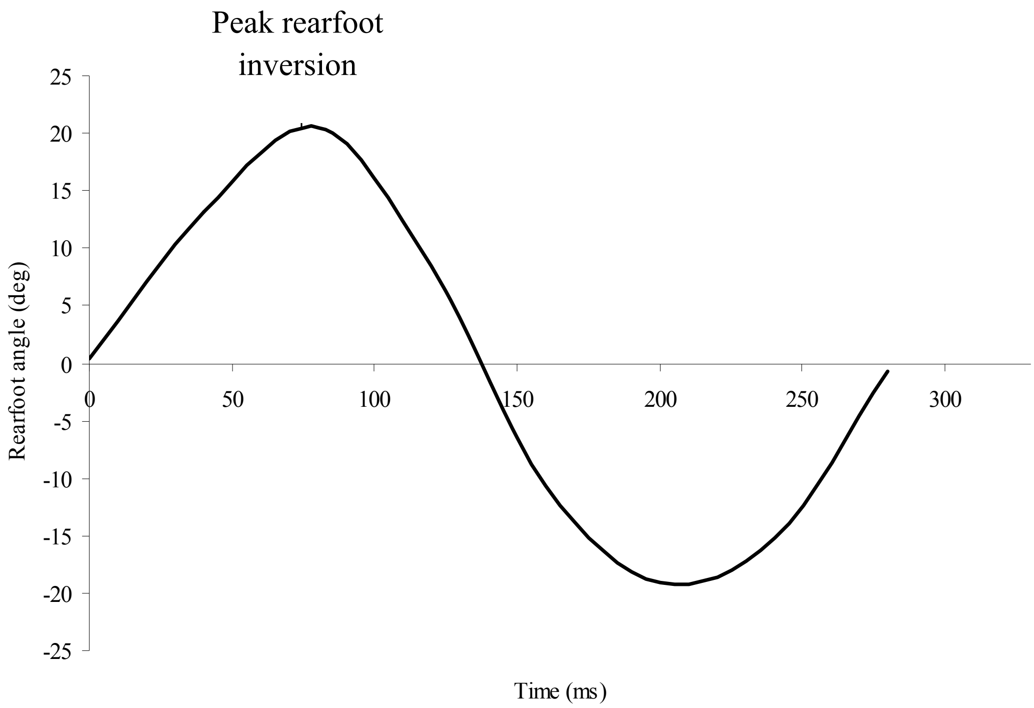


Figure 3.15. Example angle-time history showing of peak rearfoot inversion when turning

Of the kinematic observations collected during March participants exhibited significantly greater knee flexion when turning on the third generation artificial turf, compared to the natural turf whereas significantly greater inversion occurred on the natural turf (Table 3.6). The analysis showed that the footwear did not cause a change in the participants' kinematics (Table 3.7). There was also no significant interaction between the footwear and playing surface (Table 3.8).

Table 3.6 *Peak knee flexion, ankle dorsi-flexion and plantar flexion and rearfoot inversion angle data collected during March, whilst turning on the two surfaces (natural and third generation artificial turf (3g))*

	Surface	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	21.7 (13. 9)	**0.03***
	Natural	13.0 (4.9)	
Maximum Plantar flexion (deg)	3g	125.7 (9.1)	0.30
	Natural	121.3 (10.7)	
Maximum Dorsi-flexion (deg)	3g	111.8 (7.9)	0.67
	Natural	109.6 (14.8)	
Maximum Inversion (deg)	3g	15.1 (3.4)	**0.01****
	Natural	24.1 (3.9)	

Data are reported as mean and standard deviation (in brackets). Where, * denotes a significant difference at the $p<0.05$ level.

Table 3.7 *Peak knee flexion, ankle dorsi-flexion and plantar flexion and rearfoot inversion angle data collected during March whilst turning in different footwear (moulded and screw-in)*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	Moulded	21.2 (13.1)	0.70
	Screw-in	19.5 (12.7)	
Maximum Plantar flexion (deg)	Moulded	122.3 (10.8)	0.57
	Screw-in	124.7 (9.4)	
Maximum Dorsi-flexion (deg)	Moulded	108.3 (11.6)	0.34
	Screw-in	113.0 (11.7)	
Maximum Inversion (deg)	Moulded	19.8 (6.1)	0.79
	Screw-in	19.4 (5.7)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Table 3.8 *Interaction between the playing surface (natural and third generation artificial turf (3g)) and footwear (moulded and screw-in) for peak knee flexion, ankle dorsi-flexion and plantar flexion and rearfoot inversion angle data collected during March whilst turning*

	Surface	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	Moulded	27.3 (15.9)	0.57
	3g	Screw-in	28.1 (13.0)	
	Natural	Moulded	15.13 (5.8)	
	Natural	Screw-in	10.9 (2.8)	
Maximum Plantar flexion (deg)	3g	Moulded	125.5 (10.7)	0.64
	3g	Screw-in	125.9 (8.1)	
	Natural	Moulded	119.0 (10.7)	
	Natural	Screw-in	123.4 (11.2)	
Maximum Dorsi-flexion (deg)	3g	Moulded	111.4 (10.4)	0.41
	3g	Screw-in	112.0 (5.1)	
	Natural	Moulded	105.2 (12.7)	
	Natural	Screw-in	114.0 (16.5)	
Maximum Inversion (deg)	3g	Moulded	14.8 (1.4)	0.51
	3g	Screw-in	15.4 (4.8)	
	Natural	Moulded	24.9 (4.6)	
	Natural	Screw-in	23.4 (3.4)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

3.3.3. Biomechanical data (turning) - test 2 (May).

3.3.3.1. Kinetic.

Tables 3.9, 3.10 and 3.11, present the means, standard deviations and p-values for the biomechanical data taken during May whilst turning on the different surfaces, footwear variables and interactions between each, respectively.

During the second test occasion, significantly greater peak impact forces were observed on the natural turf compared to the third generation artificial surface. No significant differences were observed between the playing surfaces for the measurement of peak rate of loading, nor were there any significant differences observable for peak pressure or peak pressure loading rates (Table 3.9). No significant differences were found between footwear when measuring peak impact force or peak rate of loading. A significantly greater peak pressure was observed at the first metatarsal whilst turning in the different footwear. However, despite significant differences being observed in the main test, the post hoc tests revealed that the individual comparison of the footwear was not significant (Table 3.10). There were no significant interactions between the footwear and the playing surface (Table 3.11).

Table 3.9 *Peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected during May whilst turning on the two surfaces (natural and third generation artificial turf, (3g))*

	Surface	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	574.31 (59.60)	**0.04***
	Natural	638.94 (82.45)	
Peak rate of loading (N/ms)	3g	18.72 (5.27)	0.34
	Natural	21.60 (5.39)	
Peak medial heel pressure (H1) (N/cm^2)	3g	38.10 (17.98)	0.13
	Natural	47.56 (17.53)	
Peak lateral heel pressure (H2) (N/cm^2)	3g	29.49 (12.31)	0.4
	Natural	38.58 (14.34)	
Peak first metatarsal pressure (M5) (N/cm^2)	3g	23.86 (8.99)	0.53
	Natural	27.58 (19.09)	
Peak fifth metatarsal pressure (M1) (N/cm^2)	3g	16.75 (10.88)	0.75
	Natural	20.75 (11.19)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	1.55 (0.75)	0.27
	Natural	4.12 (7.27)	
Peak lateral heel loading rate ($N/cm^2.ms$)	3g	1.31 (0.63)	0.69
	Natural	1.46 (0.54)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	0.90 (0.44)	0.79
	Natural	1.24 (0.89)	

Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	0.59 (0.49)	0.58
	Natural	0.68 (0.50)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the p<0.05 level

Table 3.10 *Peak impact force, peak rate of loading, peak pressures and peak pressure loading rate data collected during May whilst turning in the different footwear (moulded, screw-in and soccer trainer)*

	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	Moulded	600.57 (50.93)	0.91
	Screw-in	575.56 (60.09)	
	Trainer	643.74 (104.51)	
Peak rate of loading (N/ms)	Moulded	19.58 (4.93)	0.62
	Screw-in	20.40 (8.65)	
	Trainer	20.70 (5.44)	
Peak medial heel pressure (N/cm^2)	Moulded	46.18 (16.67)	0.62
	Screw-in	35.32 (16.78)	
	Trainer	46.99 (20.27)	
Peak lateral heel pressure (N/cm^2)	Moulded	32.45 (11.41)	0.80
	Screw-in	33.92 (12.18)	
	Trainer	35.74 (18.62)	
Peak first metatarsal pressure (N/cm^2)	Moulded	17.37 (9.50)	**0.01***
	Screw-in	26.81 (9.96)	
	Trainer	12.06 (8.91)	
Peak fifth metatarsal pressure (N/cm^2)	Moulded	24.85 (8.77)	0.53
	Screw-in	29.57 (22.65)	
	Trainer	22.73 (9.70)	
Peak medial heel loading rate ($N/cm^2.ms$)	Moulded	2.02 (0.99)	0.48
	Screw-in	4.71(9.06)	
	Trainer	1.80 (0.79)	

Peak lateral heel loading rate ($N/cm^2.ms$)	Moulded	1.33 (0.55)	0.88
	Screw-in	1.40 (0.59)	
	Trainer	1.43 (0.68)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	Moulded	1.02 (0.53)	0.47
	Screw-in	1.30 (1.06)	
	Trainer	0.88 (0.35)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	Moulded	0.58 (0.41)	0.86
	Screw-in	1.01 (0.54)	
	Trainer	0.32 (0.21)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

Table 3.11 *Interaction between the playing surface (natural and third generation artificial turf, (3g)) and footwear (moulded, screw-in and soccer trainer) for the measurements of peak impact force, peak rate of loading, peak pressures and peak pressure loading rate data collected during May whilst turning*

	Playing Surface	Footwear	Mean and standard deviation (S.D)	P
Peak Impact Force (N)	3g	Moulded	589.29 (73.29)	0.12
	3g	Screw-in	539.83 (61.43)	
	3g	Trainer	593.81 (38.60)	
	Natural	Moulded	611.86 (18.46)	
	Natural	Screw-in	611.28 (35.33)	
	Natural	Trainer	693.68 (131.70)	
Peak rate of Loading (N/ms)	3g	Moulded	18.49 (4.56)	0.98
	3g	Screw-in	17.86 (6.35)	
	3g	Trainer	19.80 (6.17)	
	Natural	Moulded	20.67 (5.74)	
	Natural	Screw-in	22.95 (10.82)	
	Natural	Trainer	21.60 (5.39)	

Peak medial heel pressure (N/cm^2)	3g	Moulded	39.71 (17.49)	0.42
	3g	Screw-in	34.70 (15.66)	
	3g	Trainer	39.90 (24.71)	
	Natural	Moulded	52.66 (15.19)	
	Natural	Screw-in	35.94 (20.26)	
	Natural	Trainer	54.08 (14.62)	
Peak lateral heel pressure (N/cm^2)	3g	Moulded	31.87 (15.71)	0.37
	3g	Screw-in	29.54 (12.00)	
	3g	Trainer	27.07 (12.21)	
	Natural	Moulded	33.03 (7.45)	
	Natural	Screw-in	38.30 (12.29)	
	Natural	Trainer	44.41 (21.44)	
Peak first metatarsal pressure (N/cm^2)	3g	Moulded	16.75 (10.12)	0.75
	3g	Screw-in	24.15 (12.78)	
	3g	Trainer	9.36 (4.77)	
	Natural	Moulded	18.00 (10.35)	
	Natural	Screw-in	29.49 (7.01)	
	Natural	Trainer	14.76 (11.96)	
Peak fifth metatarsal pressure (N/cm^2)	3g	Moulded	20.56 (6.84)	0.55
	3g	Screw-in	26.15 (7.37)	
	3g	Trainer	24.86 (13.14)	
	Natural	Moulded	29.15 (9.15)	
	Natural	Screw-in	33.00 (33.34)	
	Natural	Trainer	20.59 (5.89)	
Peak medial heel loading rate (N/cm^2.ms)	3g	Moulded	1.58 (0.90)	0.55
	3g	Screw-in	1.53 (0.76)	
	3g	Trainer	1.53 (0.82)	
	Natural	Moulded	2.46 (0.99)	
	Natural	Screw-in	7.88 (0.82)	
	Natural	Trainer	2.08 (0.75)	
Peak lateral heel loading rate (N/cm^2.ms)	3g	Moulded	1.37 (0.79)	0.74
	3g	Screw-in	1.32 (0.70)	
	3g	Trainer	1.22 (0.59)	
	Natural	Moulded	1.29 (0.25)	
	Natural	Screw-in	1.47 (0.55)	
	Natural	Trainer	1.63 (0.79)	

Peak first metatarsal loading rate				
($N/cm^2.ms$)	3g	Moulded	0.61 (0.50)	0.32
	3g	Screw-in	0.86 (0.64)	
	3g	Trainer	0.32 (0.20)	
	Natural	Moulded	0.54 (0.37)	
	Natural	Screw-in	1.17 (0.45)	
	Natural	Trainer	0.33 (0.26)	
Peak fifth metatarsal loading rate				
($N/cm^2.ms$)	3g	Moulded	0.75 (0.34)	0.75
	3g	Screw-in	1.14 (0.63)	
	3g	Trainer	0.82 (0.27)	
	Natural	Moulded	1.30 (0.59)	
	Natural	Screw-in	1.47 (1.47)	
	Natural	Trainer	0.94 (0.46)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.3.3.2 Kinematic.

The kinematic data collected in May is presented in Table 3.12. A significantly greater ankle dorsi-flexion and plantar flexion was observed when turning on the third generation surface compared to the natural turf (Table 3.12). Significantly greater ankle plantar flexion was measured whilst turning in the screw-in soccer boot compared to the other footwear types (Table 3.13). No significant differences were observed for the other variables. Participants experienced no significant interactions between the surface and footwear variables whilst turning (Table 3.14).

Table 3.12 *Peak knee flexion, ankle dorsi-flexion, plantar flexion and rearfoot inversion angle data collected during May whilst turning on different surfaces (natural and third generation artificial turf, (3g))*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	20.9 (14.8)	0.93
	Natural	21.3 (11.6)	
Maximum Plantar Flexion (deg)	3g	130.5 (7.7)	**0.01****
	Natural	113.7 (22.3)	
Maximum Dorsi-flexion (deg)	3g	117.3 (9.5)	**0.01****
	Natural	98.1 (21.4)	
Maximum Inversion (deg)	3g	21.2 (6.9)	0.22
	Natural	23.7 (5.5)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the p<0.05 level

Table 3.13 *Peak knee flexion, ankle dorsi-flexion, plantar flexion and rearfoot inversion angle data collected during May whilst turning in the different footwear (moulded, screw-in and trainer)*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	Moulded	19.5 (13.9)	0.86
	Screw-in	22.6 (13.8)	
	Trainers	21.2 (12.7)	
Maximum Plantar flexion (deg)	Moulded	114.0 (24.9)	**0.02****
	Screw-in	131.9 (10.3)	
	Trainers	120.4 (13.6)	
Maximum Dorsi-flexion (deg)	Moulded	102.0 (23.7)	0.06
	Screw-in	116.6 (12.1)	
	Trainers	104.5 (17.7)	
Maximum Inversion (deg)	Moulded	20.9 (7.4)	0.3
	Screw-in	21.7 (6.0)	
	Trainers	24.7 (5.1)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the p<0.05 leve

Table 3.14 *Interactions occurring between playing surface (natural and third generation artificial turf, (3g)) and footwear (moulded, screw-in and trainer) when measuring peak knee flexion, ankle dorsi-flexion, plantar flexion and rearfoot inversion angle data during May whilst turning*

	Surface	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	Moulded	23.3 (18.0)	0.46
	3g	Screw-in	19.6 (14.0)	
	3g	Trainers	19.9 (14.8)	
	Natural	Moulded	15.8 (8.2)	
	Natural	Screw-in	25.7 (14.1)	
	Natural	Trainers	22.4 (11.5)	
Maximum Plantar flexion (deg)	3g	Moulded	127.1 (12.2)	0.12
	3g	Screw-in	133.1 (3.9)	
	3g	Trainers	131.4 (3.6)	
	Natural	Moulded	100.9 (28.2)	
	Natural	Screw-in	130.7 (14.7)	
	Natural	Trainers	109.4 (10.2)	
Maximum Dorsi-flexion (deg)	3g	Moulded	115.5 (14.3)	0.17
	3g	Screw-in	119.2 (6.9)	
	3g	Trainers	117.2 (6.9)	
	Natural	Moulded	88.5 (24.3)	
	Natural	Screw-in	113.9 (16.0)	
	Natural	Trainers	91.8 (15.9)	
Maximum Inversion (deg)	3g	Moulded	22.0 (9.1)	0.19
	3g	Screw-in	18.2 (5.9)	
	3g	Trainers	23.4 (5.2)	
	Natural	Moulded	19.8 (5.9)	
	Natural	Screw-in	25.3 (3.9)	
	Natural	Trainers	26.0 (5.0)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

3.3.4. Biomechanical data (turning) – Between test (March and May) comparison

3.3.4.1. Kinetic.

Table 3.15 presents the comparison of the turning data collected in March and May for both surfaces. On the third generation artificial turf, peak pressure and peak pressure loading rate at the medial and lateral heel and fifth metatarsal were significantly greater on the second test occasion in May compared to the first test occasion in March. When participants turned on the natural turf on the second occasion in May, peak pressure and peak pressure loading rate were significantly greater at the fifth metatarsal compared with the first test in March.

Table 3.15 *Peak impact force, peak rate of loading, and medial and lateral heel and first and fifth metatarsal peak pressure and peak pressure loading data collected whilst turning on the same playing surface (natural and third generation artificial turf, (3g)) on two occasions (March and May)*

	Surface	Test	Mean and standard deviation (S.D.)	P
Peak impact force (N)	3g	Test 1	616.07 (77.13)	0.18
	3g	Test 2	564.56 (67.96)	
	Natural	Test 1	630.75 (100.89)	0.61
	Natural	Test 2	611.57 (26.1)	
Peak rate of loading (N/ms)	3g	Test 1	19.76 (4.49)	0.52
	3g	Test 2	18.17 (5.13)	
	Natural	Test 1	18.79 (5.14)	0.39
	Natural	Test 2	21.81 (8.11)	
Peak medial heel pressure (H1) (N/cm^2)	3g	Test 1	20.27 (5.38)	**0.01***
	3g	Test 2	37.20 (15.60)	
	Natural	Test 1	31.30 (12.06)	0.12
	Natural	Test 2	44.30 (18.83)	
Peak lateral heel pressure (H2) (N/cm^2)	3g	Test 1	15.41 (3.05)	**0.01***
	3g	Test 2	30.71 (13.00)	
	Natural	Test 1	24.93 (11.47)	0.06
	Natural	Test 2	35.67 (9.84)	

Peak first metatarsal pressure (M1) (N/cm^2)	3g	Test 1	20.41 (4.56)	0.35
	3g	Test 2	23.35 (7.23)	
	Natural	Test 1	28.39 (19.78)	0.81
	Natural	Test 2	31.07 (22.72)	
Peak fifth metatarsal pressure (M5) (N/cm^2)	3g	Test 1	5.54 (2.48)	**0.01***
	3g	Test 2	20.45 (11.38)	
	Natural	Test 1	5.44 (3.37)	**0.01***
	Natural	Test 2	23.45 (10.23)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	Test 1	0.73 (1.88)	**0.01***
	3g	Test 2	1.56 (0.77)	
	Natural	Test 1	1.10 (0.49)	0.23
	Natural	Test 2	5.20 (8.90)	
Peak lateral heel loading rate ($N/cm^2.ms$)	3g	Test 1	0.52 (0.11)	**0.01***
	3g	Test 2	1.34 (0.69)	
	Natural	Test 1	1.78 (2.20)	0.78
	Natural	Test 2	1.47 (0.55)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	Test 1	0.59 (0.10)	0.08
	3g	Test 2	0.95 (0.51)	
	Natural	Test 1	0.88 (0.45)	0.23
	Natural	Test 2	1.39 (0.45)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	Test 1	0.14 (0.07)	**0.01***
	3g	Test 2	0.73 (0.55)	
	Natural	Test 1	0.14 (0.10)	**0.01***
	Natural	Test 2	0.86 (0.51)	

Data are reported as mean and standard deviation (in brackets). Where, * denotes a significant difference at the $p<0.05$ level

3.3.4.2. Kinematic.

Table 3.16 presents the kinematic data for the different surfaces between the tests occasions whilst turning. On the third generation artificial turf surface maximum inversion magnitude was greatest on the second occasion. It was also shown that peak knee flexion was significantly greater on the natural turf on the second occasion.

Table 3.16 *Peak knee flexion, ankle dorsi- and plantar flexion and rearfoot inversion angle data collected whilst turning on the same playing surface (natural and third generation artificial turf, (3g)) on two occasions (March and May)*

	Playing Surface	Test	Mean and standard deviation (S.D)	P
Maximum Knee Flexion				
(deg)	3g	1	27.7 (13.9)	0.31
	3g	2	21.4 (15.5)	
	Natural	1	13.0 (4.9)	**0.05***
	Natural	2	20.8 (12.2)	
Maximum Plantar flexion				
(deg)	3g	1	125.7 (9.1)	0.25
	3g	2	130.1 (9.2)	
	Natural	1	121.2 (10.7)	0.52
	Natural	2	115.7 (26.5)	
Maximum Dorsi-flexion				
(deg)	3g	1	111.7 (7.9)	0.16
	3g	2	117.4 (10.9)	
	Natural	1	109.6 (14.8)	0.31
	Natural	2	101.2 (23.7)	
Maximum Inversion (deg)	3g	1	15.1 (3.4)	**0.05***
	3g	2	20.1 (7.6)	
	Natural	1	24.1 (3.9)	0.43
	Natural	2	22.6 (5.6)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.3.5. Biomechanical data (running) - Test 1 (March)

3.3.5.1. Kinetic.

For each participant the force-time history occurring during each running trial was typically double peaked, in which the period up to and including the first peak indicated the impact phase (Figure 3.16).

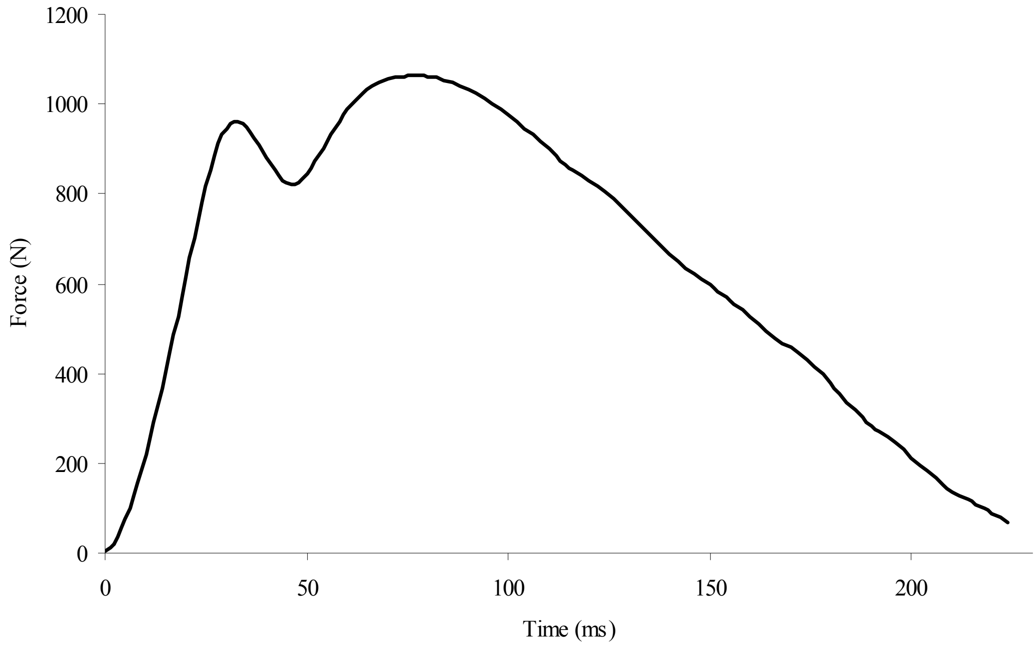


Figure 3.16. An example of a typical force-time history during running

Table 3.17 provides a comparison of the kinetic measurements taken whilst running on the natural and third generation artificial surface during March. No significant differences were observed between the artificial and natural turf surfaces for peak impact force or peak rate of loading. This was also apparent when peak pressure and peak rate of pressure loading were calculated for all masked locations.

The comparison of the kinetic measurements taken whilst wearing the different footwear conditions revealed no significant differences for any of the biomechanical variables (Table 3.18). A significant interaction was identified for peak pressure and peak pressure loading rate at the first metatarsal. Post-hoc analysis revealed that this measurement was significantly larger in the moulded boots on the third generation surface compared studded boot on the third generation surface, where as the comparison of these footwear on the natural turf was not significantly different (Table 3.19). Peak pressure loading rate also showed a significant interaction, however post hoc follow-up test with a bonforroni correction did not reveal any significant differences.

Table 3.17 *Peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected during March whilst running on two surfaces (natural and third generation artificial turf, (3g))*

	Surface	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	840.88 (127.53)	0.33
	Natural	788.36 (68.79)	
Peak rate of loading (N/ms)	3g	27.75 (11.05)	0.82
	Natural	26.62 (3.88)	
Peak medial heel pressure (H1)	3g	18.68 (9.83)	0.79
(N/cm^2)	Natural	20.40 (7.21)	
Peak lateral heel pressure (H2)	3g	17.25 (9.87)	0.53
(N/cm^2)	Natural	21.67 (9.43)	
Peak first metatarsal pressure	3g	21.25 (9.77)	0.62
(M1) (N/cm^2)	Natural	18.94 (5.13)	
Peak fifth metatarsal pressure	3g	11.11 (2.68)	0.19
(M5) (N/cm^2)	Natural	13.79 (3.22)	
Peak medial heel loading rate	3g	0.87 (0.48)	0.99
(N/cm^2.ms)	Natural	0.87 (0.36)	
Peak lateral heel loading rate	3g	0.81 (0.47)	0.58
(N/cm^2.ms)	Natural	1.05 (0.86)	
Peak first metatarsal loading rate	3g	0.45 (0.19)	0.79
(N/cm^2 .ms)	Natural	0.49 (0.17)	
Peak fifth metatarsal loading rate	3g	0.33 (0.12)	0.63
(N/cm^2.ms)	Natural	0.36 (0.11)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

Table 3.18 *Peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected during March whilst running in the two footwear conditions (moulded and screw-in)*

	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	Moulded	803.42 (93.08)	0.21
	Screw-in	825.83 (116.73)	
Peak rate of loading (N/ms)	Moulded	26.40 (9.40)	0.14
	Screw-in	27.97 (6.62)	
Peak medial heel pressure (N/cm^2)	Moulded	18.52 (9.76)	0.39
	Screw-in	20.56 (7.25)	
Peak lateral heel pressure (N/cm^2)	Moulded	18.63 (11.39)	0.50
	Screw-in	20.29 (8.09)	
Peak first metatarsal pressure (N/cm^2)	Moulded	19.64 (5.85)	0.63
	Screw-in	20.54 (9.49)	
Peak fifth metatarsal pressure (N/cm^2)	Moulded	11.95 (3.75)	0.07
	Screw-in	12.95 (2.62)	
Peak medial heel loading rate ($N/cm^2.ms$)	Moulded	0.78 (0.46)	0.22
	Screw-in	0.95 (0.36)	
Peak lateral heel loading rate ($N/cm^2.ms$)	Moulded	0.80 (0.52)	0.38
	Screw-in	1.07 (0.83)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	Moulded	0.45 (0.16)	0.54
	Screw-in	0.49 (0.20)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	Moulded	0.32 (0.13)	0.24
	Screw-in	0.37 (0.09)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Table 3.19 *Interactions between the playing surface (natural and third generation artificial turf, (3g)) and footwear (moulded and screw-in) for the measurement of peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected during March whilst running*

	Playing Surface	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	Moulded	829.91 (103.37)	0.99
	3g	Screw-in	851.86 (157.50)	
	Natural	Moulded	776.92 (82.31)	
	Natural	Screw-in	799.81 (57.62)	
Peak rate of loading (N/.ms)	3g	Moulded	25.18 (13.55)	0.07
	3g	Screw-in	30.32 (8.31)	
	Natural	Moulded	27.62 (2.74)	
	Natural	Screw-in	25.62 (4.80)	
Peak medial heel pressure (N/cm^2)	3g	Moulded	15.77 (11.06)	0.07
	3g	Screw-in	21.59 (8.37)	
	Natural	Moulded	21.26 (8.33)	
	Natural	Screw-in	19.53 (6.57)	
Peak lateral heel pressure (N/cm^2)	3g	Moulded	14.74 (12.03)	0.14
	3g	Screw-in	19.76 (7.37)	
	Natural	Moulded	22.53 (10.23)	
	Natural	Screw-in	20.82 (9.44)	
Peak first metatarsal pressure (N/cm^2)	3g	Moulded	19.42 (7.20)	**0.02***
	3g	Screw-in	23.08 (12.26)	
	Natural	Moulded	19.87 (4.84)	
	Natural	Screw-in	18.00 (5.69)	
Peak fifth metatarsal pressure (N/cm^2)	3g	Moulded	9.37 (1.19)	0.22
	3g	Screw-in	12. 85 (2.67)	
	Natural	Moulded	14.53 (3.67)	
	Natural	Screw-in	13.05 (2.83)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	Moulded	0.71 (0.56)	0.30
	3g	Screw-in	1.02 (0.35)	
	Natural	Moulded	0.86 (0.37)	
	Natural	Screw-in	0.88 (0.38)	

Peak lateral heel loading rate				
($N/cm^2.ms$)	3g	Moulded	0.66 (0.59)	0.85
	3g	Screw-in	0.97 (0.29)	
	Natural	Moulded	0.94 (0.43)	
	Natural	Screw-in	1.17 (1.19)	
Peak first metatarsal loading rate				
($N/cm^2.ms$)	3g	Moulded	0.41 (0.18)	**0.04***
	3g	Screw-in	0.49 (0.21)	
	Natural	Moulded	0.50 (0.14)	
	Natural	Screw-in	0.48 (0.21)	
Peak fifth metatarsal loading rate				
($N/cm^2.ms$)	3g	Moulded	0.26 (0.13)	0.42
	3g	Screw-in	0.40 (0.04)	
	Natural	Moulded	0.38 (0.10)	
	Natural	Screw-in	0.35 (0.12)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.3.5.2. Kinematic.

Figures 3.17, 3.18 and 3.19 illustrate the typical angle time history that occurred during running. During ground contact, participants experienced a peak knee flexion during mid-stance (Figure 3.17). During the same time, the angle-time history of the ankle angle showed plantar flexion was experienced at the beginning of the stance phase and then dorsi-flexion occurred, which peaked during mid-stance before the ankle moved into plantar flexion again (Figure 3.18). Finally, the rearfoot movement time-history showed that the participant landed in an inverted position before experiencing an eversion movement at approximately mid-stance (Figure 3.19). The rearfoot then re-inverted during the latter stages of the stance phase.

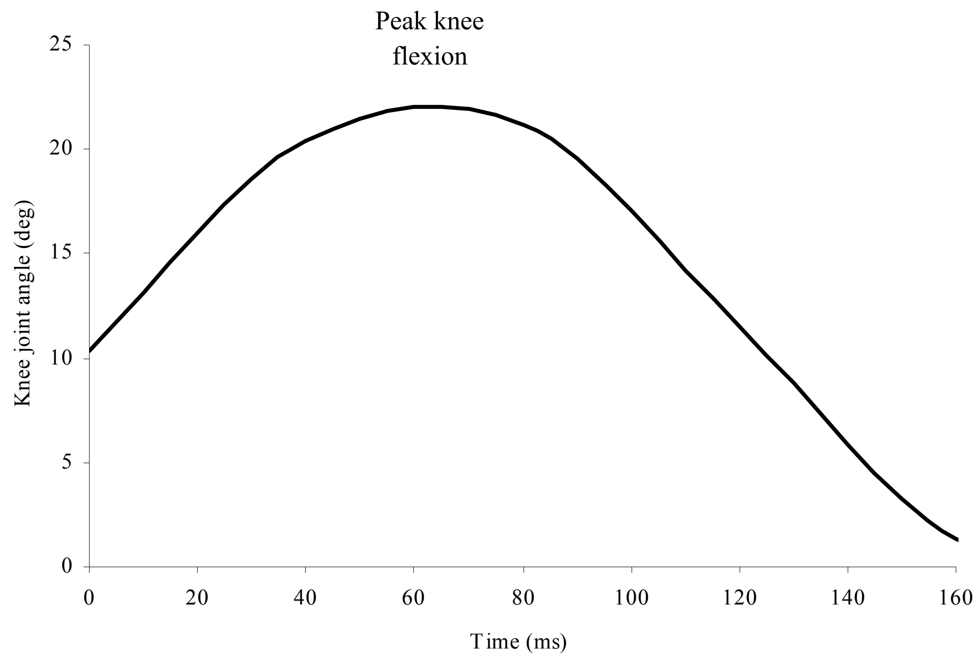


Figure 3.17. An example of a typical angle-time history showing peak knee flexion-extension when running

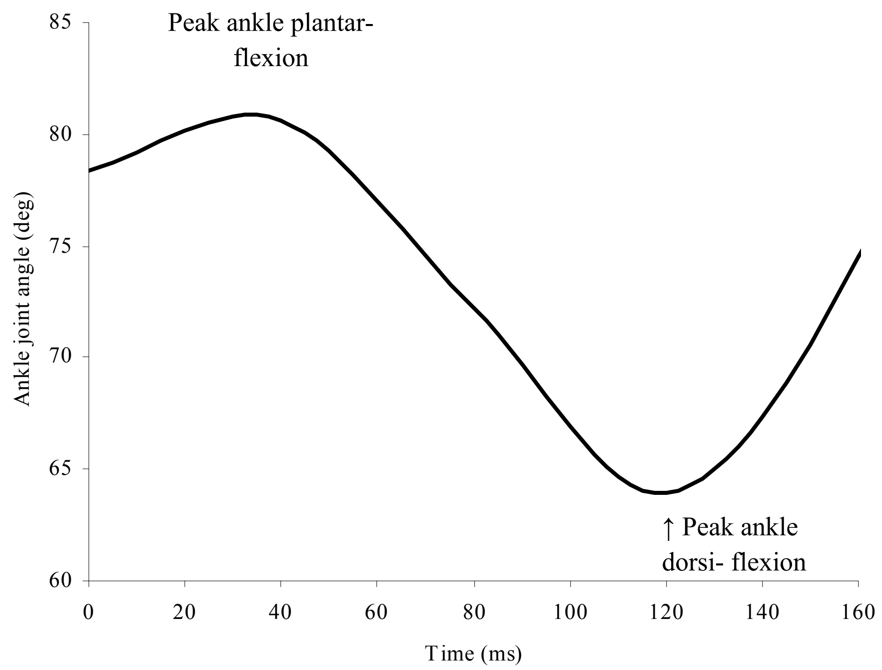


Figure 3.18. An example angle-time history showing peak ankle plantar flexion and dorsi-flexion when running

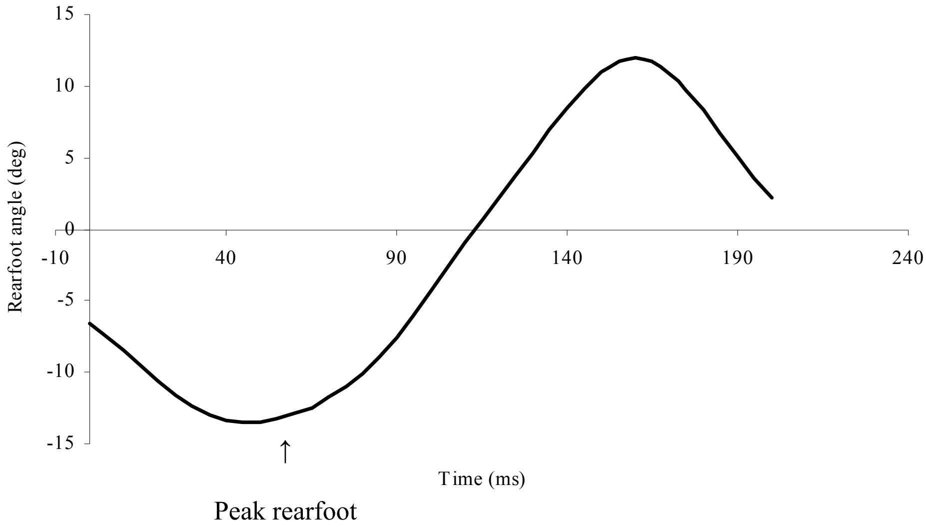


Figure 3.19. An example angle-time history showing peak rearfoot eversion when running

During test one (March) peak eversion was significantly greater on the artificial surface compared to the natural turf, whereas no significant differences were identified when comparing the peak knee flexion, plantar flexion or dorsi-flexion (Table 3.20). Likewise no significant differences were found for any measured variable whilst running in any of the footwear conditions (Table 3.21). The interactions between the footwear and surface variables were also not significant (Table 3.22).

Table 3.20 *Peak knee flexion, ankle dorsi-flexion and plantar flexion and rearfoot eversion angle data collected during March, whilst running on the different surfaces (natural and third generation artificial turf, (3g))*

	Surface	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	52.4 (12.6)	0.53
	Natural	56.5 (16.8)	
Maximum Plantar flexion (deg)	3g	98.9 (10.2)	0.3
	Natural	93.8 (12.9)	
Maximum Dorsi-flexion (deg)	3g	67.6 (11.8)	0.47
	Natural	71.7 (14.1)	
Maximum Eversion (deg)	3g	-10.9 (4.4)	**0.04***
	Natural	-6.3 (5.4)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

Table 3.21 P*eak knee flexion, ankle dorsi-flexion and plantar flexion and rearfoot eversion angle data collected during March whilst running in the different footwear (moulded, and screw-in)*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	Moulded	52.6 (15.2)	0.56
	Screw-in	56.3 (14.6)	
Maximum Plantar flexion (deg)	Moulded	98.5 (7.8)	0.39
	Screw-in	94.2 (14.6)	
Maximum Dorsi-flexion (deg)	Moulded	68.0 (9.5)	0.54
	Screw-in	71.3 (15.8)	
Maximum Eversion (deg)	Moulded	-8.13 (5.3)	0.66
	Screw-in	-9.07 (5.6)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Table 3.22 *Interactions between the playing surfaces (natural and third generation artificial turf, (3g)) and footwear (moulded, and screw-in) for the measurement of peak knee flexion, ankle dorsi- and plantar flexion and rearfoot eversion angle data collected during March whilst running*

	Surface	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	Moulded	50.7 (13.8)	0.96
	3g	Screw-in	54.1 (12.4)	
	Natural	Moulded	54.5 (17.6)	
	Natural	Screw-in	58.5 (17.5)	
Maximum Plantar flexion (deg)	3g	Moulded	100.1 (9.0)	0.70
	3g	Screw-in	97.7 (11.9)	
	Natural	Moulded	96.9 (6.8)	
	Natural	Screw-in	90.7 (17.2)	
Maximum Dorsi-flexion (deg)	3g	Moulded	67.5 (11.6)	0.57
	3g	Screw-in	67.7 (13.0)	
	Natural	Moulded	68.4 (7.9)	
	Natural	Screw-in	75.0 (18.7)	

Maximum Eversion (deg)	3g	Moulded	-10.4 (4.6)	0.97
	3g	Screw-in	-11.4 (4.6)	
	Natural	Moulded	-5.9 (5.5)	
	Natural	Screw-in	-6.8 (5.8)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

3.3.6. Biomechanical data (running) - Test 2 (May)

3.3.6.1. Kinetic.

During the second test in May, it was observed that peak vertical force could not be differentiated between the playing surfaces. Conversely, the peak rate of loading was significantly greater on the natural turf compared to the third generation artificial surface, as was peak pressure at the first metatarsal (Table 3.23). Significantly greater peak pressure loading rate was observed at the fifth at the first metatarsal in the screw-in soccer boot compared to the trainer (Table 3.24). Significant interactions were observable between the footwear and surfaces during this time for the measurement of peak pressure at the lateral heel, although post hoc analysis revealed that no significant differences could be observed for the different independent variable combinations (Table 3.25).

Table 3.23 *Peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected during May whilst running on the two surfaces (natural and third generation artificial turf, (3g))*

	Surface	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	784.51 (181.36)	0.56
	Natural	748.29 (134.82)	
Peak rate of loading (N/.ms)	3g	25.97 (4.11)	**0.05***
	Natural	32.18 (9.15)	
Peak medial heel pressure (H1) (N/cm^2)	3g	25.11 (8.48)	0.28
	Natural	33.59 (10.50)	
Peak lateral heel pressure (H2) (N/cm^2)	3g	37.64 (7.90)	0.12
	Natural	54.25 (11.25)	
Peak first metatarsal pressure (M1) (N/cm^2)	3g	29.19 (12.10)	**0.03***
	Natural	41.46 (13.34)	
Peak fifth metatarsal pressure (M5) (N/cm^2)	3g	26.91 (16.49)	0.29
	Natural	46.86 (19.93)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	1.52 (1.43)	0.92
	Natural	1.45 (0.59)	
Peak lateral heel loading rate ($N/cm^2.ms$)	3g	1.67 (0.59)	0.14
	Natural	2.72 (0.79)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	0.95 (0.58)	0.06
	Natural	1.36 (0.47)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	0.85 (0.58)	0.19
	Natural	1.6 (0.80)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

Table 3.24 *Peak impact force, peak rate of loading, peak pressure and peak pressure loading rate data collected in May whilst running in the different footwear (moulded, screw-in and soccer trainer)*

	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	Moulded	809.15 (160.71)	0.32
	Screw-in	756.61 (189.16)	
	Soccer trainer	733.43 (127.77)	
Peak rate of loading (N/.ms)	Moulded	30.44 (10.85)	0.65
	Screw-in	29.47 (5.28)	
	Soccer trainer	27.32 (6.30)	
Peak medial heel pressure (N/cm^2)	Moulded	29.48 (10.94)	0.69
	Screw-in	30.62 (9.47)	
	Soccer trainer	27.96 (11.60)	
Peak lateral heel pressure (N/cm^2)	Moulded	48.73 (17.71)	0.61
	Screw-in	43.50 (15.65)	
	Soccer trainer	45.60 (12.37)	
Peak first metatarsal pressure (N/cm^2)	Moulded	34.32 (10.52)	0.09
	Screw-in	41.16 (17.51)	
	Soccer trainer	30.48 (12.44)	
Peak fifth metatarsal pressure (N/cm^2)	Moulded	32.01 (13.59)	0.51
	Screw-in	49.66 (27.57)	
	Soccer trainer	28.78 (12.67)	
Peak medial heel loading rate ($N/cm^2.ms$)	Moulded	1.92 (1.69)	0.89
	Screw-in	1.33 (0.49)	
	Soccer trainer	1.21 (0.56)	
Peak lateral heel loading rate ($N/cm^2.ms$)	Moulded	2.48 (1.15)	0.42
	Screw-in	1.99 (0.74)	
	Soccer trainer	2.12 (0.66)	

Peak first metatarsal loading rate ($N/cm^2.ms$)	Moulded	1.06 (0.43)	0.06
	Screw-in	1.67 (1.13)	
	Soccer trainer	0.95 (0.47)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	Moulded	1.02 (0.39)	**0.03***
	Screw-in	1.58 (0.64)	
	Soccer trainer	0.86 (0.38)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the p<0.05 level

Table 3.25 *Interactions between the playing surface (natural and third generation artificial turf, (3g)) and footwear (moulded, screw-in and soccer trainer) for the measurement of peak impact force, peak rate of loading, peak pressure and peak pressure loading data collected during May whilst running*

	Playing Surface	Footwear	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	Moulded	811.62 (174.90)	0.8
	3g	Screw-in	789.70 (243.55)	
	3g	Soccer trainer	752.21 (168.20)	
	Natural	Moulded	806.69 (172.21)	
	Natural	Screw-in	723.52 (145.80)	
	Natural	Soccer trainer	714.66 (94.12)	
Peak rate of loading (N/.ms)	3g	Moulded	24.99 (4.60)	0.12
	3g	Screw-in	27.02 (5.71)	
	3g	Soccer trainer	25.92 (2.34)	
	Natural	Moulded	35.9 (13.19)	
	Natural	Screw-in	31.91 (4.05)	
	Natural	Soccer trainer	28.73 (9.04)	
Peak medial heel pressure (N/cm^2)	3g	Moulded	22.30 (5.37)	0.44
	3g	Screw-in	25.45 (4.94)	
	3g	Soccer trainer	27.59 (13.85)	
	Natural	Moulded	36.67 (10.61)	
	Natural	Screw-in	35.78 (10.66)	
	Natural	Soccer trainer	28.32 (11.03)	

Peak lateral heel pressure (N/cm^2)	3g	Moulded	34.09 (11.80)	0.10
	3g	Screw-in	35.18 (12.31)	
	3g	Soccer trainer	43.66 (17.84)	
	Natural	Moulded	63.38 (4.58)	
	Natural	Screw-in	51.82 (15.34)	
	Natural	Soccer trainer	47.56 (5.38)	
Peak first metatarsal pressure (N/cm^2)	3g	Moulded	29.94 (11.29)	0.06
	3g	Screw-in	33.66 (16.23)	
	3g	Soccer trainer	23.97 (9.09)	
	Natural	Moulded	38.71 (8.92)	
	Natural	Screw-in	48.67 (17.37)	
	Natural	Soccer trainer	36.99 (12.86)	
Peak fifth metatarsal pressure (N/cm^2)	3g	Moulded	26.56 (13.22)	0.06
	3g	Screw-in	35.20 (23.99)	
	3g	Soccer trainer	18.98 (8.41)	
	Natural	Moulded	37.86 (13.09)	
	Natural	Screw-in	64.14 (25.29)	
	Natural	Soccer trainer	38.57 (6.95)	
Peak medial heel loading rate (N/cm^2.ms)	3g	Moulded	2.17 (2.47)	0.78
	3g	Screw-in	1.19 (0.39)	
	3g	Soccer trainer	1.20 (0.57)	
	Natural	Moulded	1.67 (0.60)	
	Natural	Screw-in	1. 47 (0.60)	
	Natural	Soccer trainer	1.22 (0.64)	
Peak lateral heel loading rate (N/cm^2.ms)	3g	Moulded	1.57 (0.55)	**0.03***
	3g	Screw-in	1.56 (0.49)	
	3g	Soccer trainer	1.90 (0.79)	
	Natural	Moulded	3.39 (0.76)	
	Natural	Screw-in	2.42 (0.73)	
	Natural	Soccer trainer	2.34 (0.50)	
Peak first metatarsal loading rate (N/cm^2.ms)	3g	Moulded	0.82 (0.49)	0.75
	3g	Screw-in	1.15 (0.81)	
	3g	Soccer trainer	0.57 (0.33)	
	Natural	Moulded	1.31 (0.20)	
	Natural	Screw-in	2.18 (1.27)	

	Natural	Soccer trainer	1.32 (0.19)	
Peak fifth metatarsal loading rate (N/cm^2.ms)	3g	Moulded	0.85 (0.44)	0.08
	3g	Screw-in	1.32 (0.81)	
	3g	Soccer trainer	0.67 (0.30)	
	Natural	Moulded	1.2 (0.28)	
	Natural	Screw-in	1.84 (0.38)	
	Natural	Soccer trainer	1.06 (0.38)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.3.6.2. Kinematic.

Evaluation of the kinematic measurements taken during May showed that there were no significant differences between surface conditions whilst running (Table 3.26). There were no significant differences between the footwear conditions for any variable (Table 3.27) nor was there an interaction between the variables (Table 3.28).

Table 3.26 *Peak knee flexion, ankle dorsi- flexion, plantar flexion and rearfoot eversion angle data collected during May whilst running on the two surfaces (natural and third generation artificial turf, (3g))*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	44.8 (14.4)	0.39
	Natural	48.8 (11.2)	
Maximum Plantar flexion (deg)	3g	89.9 (10.4)	0.10
	Natural	96.1 (11.5)	
Maximum Dorsi-flexion (deg)	3g	62.1 (8.5)	0.80
	Natural	62.9 (10.8)	
Maximum eversion (deg)	3g	-12.8 (3.9)	0.17
	Natural	-14.5 (3.0)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Table 3.27 *Peak knee flexion, ankle dorsi- flexion, plantar flexion and rearfoot eversion angle data collected during May whilst running in the different footwear (moulded, screw-in and soccer trainer)*

	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	Moulded	46.6 (12.7)	0.68
	Screw-in	44.4 (9.6)	
	Soccer trainer	49.3 (16.2)	
Maximum Plantar flexion (deg)	Moulded	95.8 (11.2)	0.56
	Screw-in	91.2 (12.5)	
	Soccer trainer	92.1 (10.4)	
Maximum Dorsi-flexion (deg)	Moulded	63.1 (9.1)	0.36
	Screw-in	59.4 (8.8)	
	Trainers	65.1 (10.6)	
Maximum eversion (deg)	Moulded	-13.7 (4.1)	0.71
	Screw-in	-13.0 (3.2)	
	Soccer trainer	-14.2 (3.5)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Table 3.28 *Interaction between the surface conditions (natural and third generation artificial turf, (3g)) and footwear (moulded, screw-in and soccer trainer) for the measurement of peak knee flexion, ankle dorsi- flexion, plantar flexion and rearfoot eversion angle data collected during May whilst running*

	Surface	Footwear	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	Moulded	45.9 (15.2)	0.84
	3g	Screw-in	40.6 (9.1)	
	3g	Soccer trainer	47.9 (18.8)	
	Natural	Moulded	47.4 (10.9)	
	Natural	Screw-in	48.3 (9.2)	
	Natural	Soccer trainer	50.7 (14.8)	
Maximum Plantar flexion (deg)	3g	Moulded	95.0 (11.9)	0.27
	3g	Screw-in	83.8 (9.5)	
	3g	Soccer trainer	91.1 (7.9)	
	Natural	Moulded	96.6 (11.5)	
	Natural	Screw-in	98.6 (11.1)	
	Natural	Soccer trainer	93.1 (13.2)	
Maximum Dorsi-flexion (deg)	3g	Moulded	64.6 (7.6)	0.52
	3g	Screw-in	56.4 (7.5)	
	3g	Soccer trainer	65.4 (8.2)	
	Natural	Moulded	61.6 (10.9)	
	Natural	Screw-in	62.4 (9.5)	
	Natural	Soccer trainer	64.8 (13.4)	
Maximum eversion (deg)	3g	Moulded	-12.4 (5.7)	0.82
	3g	Screw-in	-12.0 (2.7)	
	3g	Soccer trainer	-13.9 (3.0)	
	Natural	Moulded	-15.0 (0.9)	
	Natural	Screw-in	-14.0 (3.6)	
	Natural	Soccer trainer	-14.6 (4.1)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability

3.3.7. Biomechanical data (running) – Between test (March and May) comparison

3.3.7.1. Kinetic.

The change in the playing surfaces over time was tested by comparing the surfaces between tests (Table 29). When participants performed on the third generation turf, they experienced significantly greater peak pressure at the lateral heel and the first and fifth metatarsal on the second test occasion. Participants also experienced significantly greater peak pressure loading rate at these locations. Likewise, when participants ran on the natural turf surface, they experienced significantly greater peak pressures and peak pressure loading rate at the medial and lateral heel and at the first and fifth metatarsal on the second test occasion.

Table 3.29 *Peak impact force, peak rate of loading, and medial and lateral heel and first and fifth metatarsal peak pressure and peak pressure loading data collected whilst running on the same playing surface (natural and third generation artificial turf, (3g)) on two occasions (March and May)*

	Surface	Test	Mean and standard deviation (S.D)	P
Peak impact force (N)	3g	Test 1	858.98 (154.04)	0.52
	3g	Test 2	800.66 (196.64)	
	Natural	Test 1	796.64 (75.34)	0.61
	Natural	Test 2	765.10 (154.26)	
Peak rate of loading (N/ms)	3g	Test 1	26.33 (13.11)	0.95
	3g	Test 2	26.00 (4.92)	
	Natural	Test 1	27.14 (2.82)	0.07
	Natural	Test 2	33.91 (9.28)	
Peak medial heel pressure (H1) (N/cm^2)	3g	Test 1	19.19 (10.85)	0.29
	3g	Test 2	23.88 (5.06)	
	Natural	Test 1	19.54 (6.41)	**0.01***
	Natural	Test 2	36.22 (9.86)	
Peak lateral heel pressure (H2) (N/cm^2)	3g	Test 1	17.20 (11.43)	**0.01***
	3g	Test 2	34.64 (11.18)	
	Natural	Test 1	20.20 (8.63)	**0.01***
	Natural	Test 2	57.60 (12.17)	

Peak first metatarsal pressure (M1) (N/cm^2)	3g	Test 1	21.32 (10.18)	**0.01***
	3g	Test 2	31.80 (13.09)	
	Natural	Test 1	21.83 (3.2)	**0.01***
	Natural	Test 2	43.69 (13.85)	
Peak fifth metatarsal pressure (M5) (N/cm^2)	3g	Test 1	10.20 (1.95)	**0.01***
	3g	Test 2	30.88 (18.52)	
	Natural	Test 1	13.57 (3.14)	**0.01***
	Natural	Test 2	51.00 (23.34)	
Peak medial heel loading rate ($N/cm^2.ms$)	3g	Test 1	0.89 (0.53)	0.24
	3g	Test 2	0.88 (0.34)	
	Natural	Test 1	1.68 (1.72)	**0.01***
	Natural	Test 2	1.57 (0.57)	
Peak lateral heel loading rate ($N/cm^2.ms$)	3g	Test 1	0.83 (0.55)	**0.01***
	3g	Test 2	1.56 (0.48)	
	Natural	Test 1	1.10 (1.02)	**0.02***
	Natural	Test 2	2.90 (0.86)	
Peak first metatarsal loading rate ($N/cm^2.ms$)	3g	Test 1	0.42 (0.15)	**0.01***
	3g	Test 2	1.08 (0.65)	
	Natural	Test 1	0.55 (0.14)	**0.01***
	Natural	Test 2	1.52 (0.46)	
Peak fifth metatarsal loading rate ($N/cm^2.ms$)	3g	Test 1	0.29 (0.13)	**0.01***
	3g	Test 2	0.99 (0.64)	
	Natural	Test 1	0.37 (0.10)	**0.01***
	Natural	Test 2	1.74 (0.96)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.3.7.2. Kinematic.

Table 3.30 presents the kinematic data for participants running on the same turf on two different occasions. It is observable that participants experienced significantly greater plantar flexion on the third generation turf on the first test (March) compared to the second (May). Participants also experienced significantly greater rearfoot eversion on the natural turf on the second test occasion than on the first.

Table 3.30 *Peak knee flexion, ankle dorsi- and plantar flexion and rearfoot inversion angle data collected whilst running on the same playing surface (natural and third generation artificial turf, (3g)) on two occasions (March and May)*

	Playing Surface	Test	Mean and standard deviation (S.D)	P
Maximum Knee Flexion (deg)	3g	1	52.4 (12.6)	0.09
	3g	2	43.3 (12.3)	
	Natural	1	56.5 (16.8)	0.14
	Natural	2	47.8 (9.6)	
Maximum Plantar flexion (deg)	3g	1	98.9 (10.2)	**0.05***
	3g	2	89.4 (11.8)	
	Natural	1	93.8 (12.9)	0.44
	Natural	2	97.6 (10.8)	
Maximum Dorsi-flexion (deg)	3g	1	67.6 (11.8)	0.10
	3g	2	60.5 (8.4)	
	Natural	1	71.7 (14.1)	0.06
	Natural	2	62.0 (9.7)	
Maximum eversion (deg)	3g	1	-10.9 (4.4)	0.45
	3g	2	-12.2 (4.2)	
	Natural	1	-6.3 (5.4)	**0.01***
	Natural	2	-14.5 (2.6)	

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

3.4. Discussion

The current study has highlighted differences between natural turf and third generation surfaces at different times of the year. Likewise differences were found between footwear and between the same natural turf and third generation surfaces at different times of the year. The reasons for the differences and the implications of these finding are discussed below.

3.4.1. Third generation surfaces and natural turf.

The observation that during the complete 180° turning movement, peak inversion magnitude was significantly greater on the natural surface with the potentially lowest traction, is counter-intuitive and in contrast to situations where lateral ankle damage are common (Orchard, 2002; Orchard & Powell, 2003). It is suggested that the type of movement performed could have influenced the turning strategy. During the turn, the right foot acted as a pivot whilst helping to propel the body forwards, whereas the left leg was used as a support. When the surface traction was low, the right foot was able to slide as the resistance to the foot is small. Therefore, whilst the left leg and body was fixed during the turn, the right leg slid causing the leg to extend. During this extension of the leg, the plantar studs were used to grip the surface in order to create traction. This however, may have caused a larger relative rearfoot angle to be experienced. This notion is supported by the kinematic observation that a significantly reduced maximum knee flexion was observed on the natural surface suggesting that the leg may have stayed in an extended position for longer. The magnitude of the knee flexion on the artificial surface may also be related to the role of increased knee flexion as a mechanism by which traction could be lowered (Durá et al., 1999). A combination of these factors may have resulted in the significant differences in knee flexion being observed.

The rearfoot inversion values in May showed different patterns to those observed in March. Despite a significant increase in plantar flexion on the third generation surface, which was suspected to allow for accentuated rearfoot inversion, no significant difference in rearfoot inversion was observed. Instead, it is suggested that the lower plantar flexion on the surface with the higher traction (natural turf) is a protection mechanism to prevent excessive inversion, by reducing the range of possible inversion.

The assessment of the surfaces using measurements associated with overuse injury also revealed interesting results. Prior mechanical testing of the surfaces in March indicated that cushioning differences were evident between the surfaces, where the natural turf possessed greater cushioning than the third generation turf. Despite this, peak vertical forces during running and turning were not significantly different. This however, is consistent with previous literature comparing forces whilst running on concrete, asphalt, acrylic and natural sports surfaces (Dixon & Stiles, 2003; Dixon et al., 2000; Dixon et al., 2008).

One reason for the lack of significant differences may be that the peak impact force measurement is not sensitive to changes in surface cushioning (Hamill, 1996). Instead, researchers have advocated the use of peak rate of impact loading to observe differences between surfaces (Dixon et al., 2008; Dixon & Stiles, 2003). The data provided for running or turning in March fail to support such a suggestion. It is possible that the participants changed their lower extremity geometry to lower the loading experienced (Ferris et al. 1998; Ferris et al., 1999). This suggestion is supported by the observation that participants exhibited a significant increase in the peak rearfoot eversion in response to the mechanically harder third generation artificial turf surface. This is a primary method by which forces can be reduced (de Wit et al., 1995; Stacoff et al., 1988; Whiting & Zernicke, 1998) and possibly indicates why when running no differences in force variables were apparent. In contrast and as explained earlier, during the performance of the turning movement, maximum rearfoot inversion increased on the more cushioned surface, which is somewhat difficult to explain. In theory, this additional inversion would also act as an impact attenuation mechanism during lateral movements (Dayakidis & Boudolos, 2006), yet this was not reflected in the relative magnitudes in rearfoot inversion between the surfaces. It is possible that the more cushioned surface was more unstable and thus caused greater rearfoot inversion.

A problem with the measurement of impact force and peak loading rate is that for both measurements it is assumed that the forces measured are located at the heel, detailing the magnitude of the shock waves travelling directly vertically through the segments (Clingham et al., 2008). This however, has been shown not to be the case (Shorten, 2002). True, most force generated at impact is at the heel. Resultant impact force however, is the summation of heel as well as some mid- and fore-foot force (Shorten, 2002) and during running the forefoot comes into contact with the ground at around 8% of the stance phase, which is approximately 20 ms (De Cock et al., 2005) and well within the 50 ms defined as the impact phase (Dixon et al., 2008). Thus, identifying heel forces or pressure in isolation may provide a suitable method for comparing shoe and surface conditions (Dixon et al., 2008) and may offer greater insight and sensitivity to surface changes than measures of resultant forces (Dixon et al., 2008; Tillman et al., 2002; Dixon & Stiles, 2003). Such theory is supported in the current investigation, whereby increased lateral heel pressure is experienced when turning on the third generation turf, the harder of the two surfaces. In contrast, medial heel peak pressure did not change and peak pressure loading rate at this location was greatest on the natural turf. Expecting that the increased surface cushioning would have a uniform effect on reducing all peak pressures, the primary mechanism by which the change in pressure was evident was more likely a consequence of the foot position rather than the cushioning of the surface. This replicates those findings of Ford et al. (2006), and possibly suggests that the surfaces were not distinct enough to elicit biomechanical cushioning to all regions. Never-the-less, despite a direct relationship between the pressure patterns experienced and specific overuse injuries and stress fractures being difficult to establish (Eils et al., 2004) it is imaginable that pressure patterns during running and turning movements may lead to the typical stress fractures experienced in soccer when coinciding with high repetition and inadequate rehabilitation time (Eils et al., 2004). This may well also result in a perceived reduction in comfort which may change the gait of the participant further, influencing the loading of the participant (Wong Chamari, Mao, Wisløff & Hong, 2006). By observing that these differences in pressure were found during turning and not running, this indicates the potential role of the movement to cause different loads and change the injury risk otherwise not experienced during running, and highlights the importance of selecting a range of appropriate movements when testing playing surfaces.

The observation that these measurements of loading were different on the alternative surfaces, may indicate the potential change in injury risk, whilst also highlighting the ability of the pressure data to detect differences between surfaces that other measurements such as force plate data have not. This is consistent with past studies (Dixon et al., 2008; Dixon & Stiles, 2003; Tilmann et al., 2002). Although lateral heel pressure measurements were increased on the third generation turf which is consistent with an increase in lower extremity loading, the observation that lower medial heel peak pressure loading rate was evident on the same surface indicates that across this specific region the loading may have been reduced. These observed significant differences between conditions at the medial and lateral heel however, correspond with differences published by Ford and colleagues (2006), albeit of the forefoot and not the heel. Yet, these authors reported that peak pressures were greatest on the medial forefoot on the natural turf and greatest on the lateral forefoot on the third generation surface. Despite this trend between pressure patterns and surfaces between the studies, the findings are also contrasting, as Ford and co-workers suggested that the artificial surface would be more cushioned than the hard natural turf, whereas the opposite was reported in the current comparison of these surfaces in March. The increased medial pressure observed by Ford et al (2006) was suggested to result in greater rotation traction generated during propulsion of cutting movements on natural turf, whereas the increased lateral heel pressure may suggest a change in linear traction on the third generation turf.

Pressure measurement and force calculation at the foot–shoe interface has been described as occurring as a consequence, in part, of the loading patterns at the boundary between the shoe and surface (Ford et al., 2006). Nigg and Segesser (1992) cite evidence to suggest that the shoe–surface interface is an important variable related to injury frequency and severity, and in particular high frictional conditions at the shoe–surface interface may contribute to higher injury rates. Finally, Cawley, Heidt Jr, Scranton Jr, Losse and Howard (2003) found elevated frictional resistance of 30–1500% at the ground surface when the axial load was increased by 400%. As a consequence, the higher loaded foot regions may produce higher frictional resistance at the shoe–surface interface than the areas of lower loads (Ford et al., 2006), the foot may then 'stick' during linear movements and the forward momentum of the body may cause the ankle to roll, potentially damaging the ankle ligaments (Blazevich, 2004).

Pressure data may also be associated with the kinematic data. During the turning movement, the nature of the task requires the participant to land on the medial side as the rearfoot inverts. However, as the level of foot slide increases, movement of the calcaneous may not occur to the same extent. With moderate levels of slide, the extension of the shank may have been enough to show the increased total rearfoot angulations observed in March. This may have occurred due to the surface not causing substantial foot slide or it may result from increased muscle action from the medial ankle to allow greater movement of the calcaneous to provide surface grip. This suggestion is supported by the reduced lateral pressures on the natural turf while maximum inversion values were significantly lower than on this surface. With greater movement speeds or a change in surface conditions, more rapid slides are possible. In such situations it is possible that the movement of the calcaneous may become even less, potentially causing further reduction in lateral heel pressure. In this situation the inversion angle decreases and slipping becomes more likely. Conversely, on the third generation surface, the high pressure suggests increased traction and at increased speeds a greater momentum of the body may cause ankle roll to damage the lateral ankle structures.

In contrast to the data presented in March, no difference was shown between the medial and lateral heel pressures or pressure loading rates between surfaces in May during either running or turning, despite the environmental conditions being more similar to those presented by Ford et al (2008). Significantly greater peak pressures however, were observed at the first metatarsal when running on the natural turf, which would indicate the possibility of an increased injury risk to this metatarsal with the use of the third generation artificial surface. This also shows greater medial loading of the fore-foot as was shown by Ford et al. (2006). Failing to show changes in the other pressure measurements however, suggests that this change may be related to the change in foot position. It may however, also reflect the reduced statistical power evident due to participant drop out (N= 4) in this comparison. The difference at the first metatarsal location may relate to the turning movement and the greater loading on this side of the foot in response to the dynamic movement. The load on this structure is greater during turning than during running. As a consequence the cushioning may have a larger effect at this location and thus the difference may be more observable significantly (due to a larger effect size).

Whilst running, consistent with the findings in March, the surfaces in May were not distinguishable when using peak impact force. However, the lower value for peak loading rate on the natural turf when running suggests a change in the lower extremity loading on this surface compared to a dry natural turf. This would indicate potential injury reduction if the artificial surface was used when natural turf becomes hard (i.e. preseason). During turning, impact forces were also significantly lower on the third generation turf. Observing that by using a more dynamic movement, impact force measurements were able to distinguish between surfaces is replicable of findings presented by Stiles and Dixon (2006), although they reported that peak impact force values were lower on the least cushioned surface. No kinematic changes were evident in the present investigation which may account for the change in force. Possible reasoning behind the observation of differences during turning and not running may be that impact force, as explained earlier, is the summation of the forces during the impact phase which includes forces from the heel and some from the mid- and fore-foot (Shorten, 2002). During cutting (a not dissimilar movement to turning), greater medial mid- and fore-foot peak pressure have been found to occur compared to running whilst lateral heel pressures were not different (Eils et al., 2004; Wong et al., 2006). Consequently, assuming that this force is also produced within the first 50 ms, the larger force production from the mid- and fore-foot would contribute to a greater extent to the magnitude of impact force than during running.

The results of this investigation show that lower extremity loading of the soccer players were shown to be somewhat influenced by the use of third generation artificial surface. However, as the construction of the artificial turf can differ, particularly the shock pad density provided beneath the top surface layer, the construction may influence the response of the athlete and differences may be observed. It is suggested that these differences be quantified.

In summary, playing surfaces did alter the loading of the lower extremity but this may depend on the time of year and the movement performed. Some of the plantar pressure measurements were able to distinguish between surfaces whereas others could not. In such cases where pressure measurements were different, changes in foot positions in March were more likely to be the cause of measurement change rather than as a direct response to the surfaces. This may explain why some areas experienced change and others did not. These kinematic changes are likely to result from the perception of the surface. In May, peak rate of loading and peak pressure at the first metatarsal during running and peak impact force during turning were able to differentiate between surfaces. This highlights the importance of selecting different movements and suggests that during summer when surfaces are hard and dry, third generation surfaces may prove useful for injury prevention. However, the observation of significantly increased peak impact force on natural turf may have resulted from increased mid and fore-foot load and may not reflect a change in load through the heel.

3.4.2. Soccer boots.

The observation that during March none of the biomechanical measurements associated with lateral ankle injury were significantly different across footwear conditions, indicates that the risk of injury did not change at this time with the different boots. It also supports the earlier suggestion that the turning speed may not have been sufficient to elicit significant differences and that protection may be provided by additional muscular action. Similarly, the finding that peak pressure measurements were not significantly different between the footwear when running or turning in March supports the results of Coyles and Lake (1999). These authors assumed that this unchanged pressure was a consequence of the similarities in stud location on the outsoles. It was also explained that the combined influence of both sufficient stud number and outsole plate stiffness may have been adequate to distribute the pressure across the whole of the forefoot, as predicted by Shorten (1998). Further still, the studs may have also been able to penetrate the surface fully so that the sole plate of the boot made contact with the surface (Kirk et al., 2007). This may have also enabled similar peak pressure and peak pressure loading rates to be shown. Impact force and peak loading rate were not significantly different between footwear, which is also consistent with Dixon et al. (2008), when moulded and screw-in soccer boots were compared.

By contrast footwear differences were evident in May, where significantly greater plantar flexion occurred when turning in the screw-in soccer boots. Increased plantar flexion has been associated with increased strain on the ATFL (Fujii et al., 2005) and was experienced in screw-in boots compared to moulded boots and trainers. Maximum inversion however, did not differ between the boots. Thus the increased plantar flexion only makes the foot more susceptible to excessive movements, but does not induce greater instability (Fujii et al., 2005). It is possible that increased muscular control is applied which enables less movement, and therefore protects the ankle (Hopkins et al., 2007). It may take the experience of an unexpected factor such as a sudden change in surface type and/or consistency, an awkward foot plant, or an increase in movement speed before the magnitude of the inversion movement increases significantly.

The observation of no significant difference in impact forces or loading rates between footwear conditions is consistent with Dixon et al. (2008), yet participants did experience significantly greater peak pressure loading rate on the fifth metatarsal in the screw-in studded boot condition during running compared to the soccer trainer. This may have occurred because the screw-in stud is long and metal ended, with a stiff sole plate, whilst soccer trainers do not have any studs, and have more mid-sole cushioning (Queen et al., 2008). Thus, this additional cushioning provided by the soccer trainers may have delayed the loading of the pressure experienced by the participant. This would support the suggestion of Queen et al. (2008) that the use of soccer trainers may enable a reduction in metatarsal injury.

It was found that when turning in March the measurements of peak pressure loading rate at the first metatarsal were greater in the studded boot compared to the moulded on the third generation artificial turf when performing a turning movement. No differences were observed between the footwear on the natural turf. Observing only a change in peak pressure at the first metatarsal and not the other locations may suggest that kinematic changes are likely to be responsible for such trends, although observing no significant kinematic changes makes explaining such findings difficult. When running in March, analysis revealed that the first metatarsal peak pressure measurement was significantly larger in the moulded boots on the third generation surface compared screw-in boot on the third generation surface, whereas the comparison of these footwear on the natural turf was not significantly different. As such, the choice of surfaces on which to test footwear seems to be an important characteristic of any research investigation, which replicates previous opinion (Dixon & Stiles, 2003; Dixon et al., 2008) and indicates that loading differences can be observed given proper consideration of the surface conditions. Dixon et al. (2008) however, observed significant reductions in peak rate of loading when running in the moulded soccer boot compared to the screw-in soccer boot, which was not evident here. The reason for this may be that the running speeds and boot designs may have been sufficiently different between studies to load the foot in different ways. Likewise, the provision of artificial grass to run on may have changed the movement of the participants in the current investigation compared to those used by Dixon and colleagues, thus affecting the relative difference between the soccer boots to a different extent. Further still, having observed that significantly greater peak pressure and peak pressure loading rate were experienced, loading does appear to be changed although this depended on the movement type. As such, further emphasis is provided regarding the importance of selecting a range of movements to perform in research investigations, conclusions consistent with Queen et al. (2008)

In summary, the role of the studded soccer boot designs (moulded and screw-in) in changing the biomechanical responses of soccer players appears somewhat limited. In situations where significant differences were shown, findings were dependant on the surface condition and movements, and thus indicate that when investigating footwear thorough consideration of the surface conditions and movements are needed.

3.4.3. Natural turf surfaces: test one and two.

When comparing the biomechanical measurements associated with overuse injuries, it was hypothesised that the magnitude of lower extremity loading would be significantly increased when running and turning on the natural surface when the environment was warm and dry. The mechanical tests reported differences in the cushioning of the natural turf between the two test occasions, yet peak impact forces were no different whilst turning or running, again replicating trends in the literature concerned with playing surfaces (Dixon et al., 2008; Dixon et al., 2000; Dixon & Stiles 2003), and in particular those trends presented by both Dixon et al. (2008) and Stiles et al. (2006), who tested participants whilst running on natural soil surfaces that were distinct in their mechanical cushioning. Peak rate of loading was also not significantly different when running or turning on the different surfaces, which is also true of the findings presented by Dixon and colleagues (2008) on contrasting soil densities, but are in contrast to Stiles et al. (2006). Increased rearfoot eversion may account for the lack of differences during running, although it is unclear why values are also similar during turning.

Having previously advocated the improved ability of in-shoe pressure data to detect differences in surface cushioning, the observation that plantar pressure and peak pressure loading rate, this time at the medial and lateral heel and the first and fifth metatarsal, were distinguishable between surfaces whilst running adds further support to this. In this comparison it seems likely that the changes in values are not in response to a change in foot position but rather a response to the difference in surface cushioning. The uniform decrease in pressure measurements indicates that increased surface deformation increases the contact area between the shoe sole and surface thereby causing a redistribution of the localised forces (Hennig et al., 1996; Le Veau, 1992; Stiles & Dixon, 2007). This finding is also comparable to those presented by Dixon et al. (2008) when testing 'hard' and 'soft' soil in a soil bin. This finding may indicate a reason for injury on the harder surface and a possible reason for an increased risk of injury during preseason. Finding that during turning only peak pressure and peak pressure loading rate at the fifth metatarsal was different, could relate to the small sample size used to make the comparison and thus a lower statistical power. Having observed significantly increased kinetic values when the surface was dry compared to when the surface was wet, it is suggested that watering of hard surfaces may provide a method by which biomechanical measurements associated with overuse injury could be reduced. Associated measurements related to acute injury were not significantly different suggesting no change in injury risk between the test sessions, which is in contrast to injury studies which have demonstrated that acute injuries have occurred under similar conditions (Orchard, 2002; Orchard & Powell, 2001; Woods et al., 2003), yet may be linked with test protocol, where the speed of the movement and awareness of the task may have enabled the participant to use muscular support to prevent excessive movement of the joint.

3.4.4. Third generation artificial surface: test one and two.

The suspected consistency of the third generation turf surface over time (Veenbrink, 2002) is questioned by the evidence that peak pressures and peak rate of pressure loading were significantly different between test sessions. Unlike peak impact forces and peak rate of impact loading that were not significantly different, it was shown that peak pressure and peak pressure loading rate at the fifth metatarsal and at the medial and lateral heel were greater on the second occasion when turning. In addition, peak rearfoot inversion was greater at this time. Likewise, whilst running, peak pressure and peak pressure loading rate at the first and fifth metatarsal and lateral heel were significantly greater on the second occasion. This supports the earlier speculation that pressure measurements are sensitive to small changes in surface cushioning. The lack of statistical power indicated in the previous section may also account for the non-significant difference in peak pressure and peak pressure loading rate at the first metatarsal and medial heel, found when turning and running respectively.

The biomechanical changes that occurred between March and May may have resulted because of wear related damage from overuse, the type of third generation construction, the quality of installation, and length of time between tests. These factors may all influence the consistency of the surface and thus the biomechanical response. Previous investigations using mechanical tests have assessed the effect of environmental conditions, particularly the effect of surface moisture, on the properties of third generation surfaces. They found that environmental conditions did not influence the impact attenuation provided by the surface (McNitt et al., 2004). Mechanical tests have shown that the condition or damage of the rubber can influence the force reduction properties (Sifters & Beard, 1994), although this was assumed to occur over long periods of time (Fleming et al., 2008). Degradation in mechanical cushioning on the third generation artificial surface can result from ineffective drainage of the surface water. This, as well as the collection of other substances seeping into the surface, may affect the ability of the rubber crumb to be displaced, impairing the effectiveness of the surface to dissipate the force (Popke, 2002). The loss in rubber crumb may also reduce the contact area between the foot and the surface, producing significantly greater pressures.

The data of the current investigation highlight the improved measurement sensitivity of biomechanical testing, and thus importance of using human participants to test a surface compared to mechanical tests. Results also indicate a more rapid change in surface condition than first thought and that mechanical tests are unable to detect such a change. Thus it is suggested that improved mechanical tests are needed that are used more frequently throughout a season.

3.4.5. Study limitations.

Care must be taken when interpreting the results of the kinematic observations in the current study. As the cameras were not mechanically synchronised, the reconstruction of the joint coordinates will consequently include additional error. This error was limited in the first instance by matching the cameras using a distinct event. In this case the moment of initial ground/heel contact was used. The remaining error was then estimated by calculating the average difference in peak angle that occurred across eight trials for both running and turning. Results from this sensitivity analysis indicated that the maximum error varied depending on the movement and angle calculated. However, the range was not significant and thus confidence in the differences between the independent variables is high. However, because the collection frequency for the kinematic data was only 25 Hz, it is possible that 'real' peak values may have been missed, particularly those measured in the early part of the stance phase.

Another limitation of the study is that only sagittal and frontal plane joint kinematics were calculated. The three dimensional reconstruction of the marker co-ordinates allowed each marker to be reconstructed whilst removing the different perspective and parallax errors associated with two dimensional data collected with one camera. Despite this, however, all of the kinematic measurements were measured in the frontal or sagittal planes and this has been shown to introduce error in angular calculations, particularly for the frontal plane (Soutas-Little et al., 1988). In order to improve the accuracy of the kinematic analysis, construction of local reference planes defining the actual movement plane enabling three-dimensional analysis of peak ankle, knee and rearfoot movement would improve this limitation and would offer improvement to the measurement accuracy. Further still, the software used to reconstruct the marker coordinates required manual digitisation where accuracy depends on the digitiser's ability and experience. Thus improvements to the method could also be made if an automatic tracking system was used.

Further consideration should be given to the measurement of peak forces. In this study the forces were collected via the pressure insoles placed into the footwear of the participant. This allowed the plantar foot pressures to be measured, which do not provide a true reflection of the vertical force. The sensors that are used to calculate the resultant force occurring at impact, measure the normal force; that is, the force that occurs in a perpendicular direction to the sensor (Kalpen & Seitz, 1994). Therefore, any force that occurs in any other direction to the sensor is not measured. In addition, there is 'dead space' located between each sensor that does not measure vertical force (Kalpen & Seitz, 1994) which can result in an underestimation of the force and pressure that occurs.

The turning movement has been shown to produce reliable pressure and force data. However, in response to the turning manoeuvre, the participant may influence the turning strategy through changes made to the step prior to the turn for different conditions. As such, a measurement of this alteration would benefit the knowledge of turning movement characteristic.

Previous observations have shown that the risk of injury is higher in preseason and early in-season compared to the rest of the year (Hawkins et al., 2001; Woods et al., 2002; 2003). However, in the current study, the times tested corresponded to periods of the year where the conditions were contrasting in order to provide some insight in to how the environment influences the relative comparison different surfaces and footwear. It is a limitation of this study that comparisons were not made during pre-season and other in-season periods, but time constraints and bad weather made testing at these periods difficult. This however, is worth considering in future research to confirm the findings presented in the current study, to improve the understanding of the mechanisms behind injury and the different responses that occur on third generation artificial surfaces and natural turf at different times of the year.

Finally, this study had experimental design limitations. One such limitation to the study was that data were only collected on amateur level soccer players and thus may not be generalised across different sporting populations (Ford et al., 2006). In addition, participants were not blind to the surfaces and therefore the perceived condition of the surface and expected outcomes could not be controlled (Ford et al., 2006).

3.4.6. Conclusions

Drawing on the evidence of the current investigation, there is some indication of lower extremity loading change through significant biomechanical differences between the natural and third generation surfaces at different times of year. Having shown that the rearfoot inversion measurements associated with acute injury increased on the natural surface which was suspected to have lowest traction during March and that no differences occurred in May, the hypotheses that rearfoot inversion would increase on the third generation turf in March and natural turf in May must be rejected. Observing increased peak pressures on the third generation turf compared to the natural turf on the first test occasion indicates that the hypothesis that lower extremity loading would be reduced on the natural turf was at least partly accepted. Likewise the finding that peak impact force and peak pressure during turning and peak rate of loading during running were lowest on the third generation turf in May is in agreement with the hypothesis that lower extremity loading would be greater on the natural turf during the second test occasion. Given these findings, consideration should be given to the range of environmental conditions in which the playing surface should be tested influencing conclusions of research papers. Without a sufficient range of conditions, comparisons are only appropriate to the time of year that the surface is tested. Therefore, not only do biomechanical investigations require the appropriate surface condition to test footwear (Dixon et al., 2008; Queen et al., 2008), which was also supported by the evidence presented in the current chapter, the environmental conditions are also an important consideration when assessing different surfaces.

What was also made clear by the evidence presented was that certain biomechanical measurements of lower extremity loading are more sensitive to changes in surface than others. By using plantar pressures, differentiation could be made between natural and third generation artificial surface conditions where resultant force measurements could not. Owing to their specific indication of heel force, these measurements are better suited to indicating lower extremity loading and thus more able to distinguish surfaces that have been shown to be associated with injury risk. The evidence also emphasised the importance of selecting different movements, as the observation of biomechanical differences was dependent on the test movement.

Studded soccer boots (moulded and screw-in) have very little cushioning and were often indistinguishable from one another when biomechanical data were presented, particularly on natural turf surfaces. This was in contrast to the hypothesis that the magnitude of inversion would be significantly greater in the longer screw-in studs compared to the moulded boots and the hypothesis that lower extremity loading would be increased n the longer screw-in studs compared to the moulded boots. Future research should therefore focus on methods in which soccer footwear can be designed to provide greater attenuation of loads during locomotion, particularly at the heel. Cushioning provided by the soccer trainer was only significantly lower at the fifth metatarsal compared to the screw-in soccer boot, so the hypothesis concerned with reduced lower extremity loading in the soccer trainer can be partly supported.

Environmental conditions influence surfaces cushioning, moisture content, and surface temperature. These factors have been associated with the magnitude of traction provided to the soccer player and increased risk of experiencing inversion injury (Orchard, 2002; Orchard & Powell, 2003; Torg et al., 1996). It was therefore hypothesised that biomechanical values associated with acute injury on natural turf would be greatest during May when the surface was harder compared with March. In light of the fact that there were no significant differences shown for the measurements associated with acute injury, this hypothesis was rejected. It was also hypothesised that biomechanical values associated with overuse injury would be greater on the natural turf in May compared with March. Since plantar foot pressure loading was greater on the second occasion, this hypothesis was supported.

Finally, due to the potential of the artificial surface to remain consistent for long periods of time, it was hypothesised that lower extremity loading and movements associated with acute injury would not differ between the artificial surfaces at the different times of the year. Observing that pressure data and rearfoot inversion did differ, this hypothesis cannot be supported.

In conclusion, differences between natural and third generation artificial surfaces were evident, although the differences depended upon the time of year and the movement performed. Likewise, biomechanical differences were evident for both 3g and natural turf surfaces between the two tests. As such, this suggests that both of these surfaces are influenced by the environmental conditions. The construction of the soccer boots used (screw-in, moulded) seemed to have little influence over the loading of the lower extremity during running and turning, although if interactions with playing surface were taken into consideration, differences are observed. Finally, both the natural and third generation turf surfaces showed changes over time, with increased biomechanical values, indicating a change in loading.

4. Study Two:
The Influence of Shock Pad Density and Soccer Boot Cushioning on Biomechanical Measurements

4.1. Introduction

As previously highlighted, soccer is traditionally played on natural turf surfaces (Ekstrand et al., 2006). However, the suitability of these surfaces for use in match and practice situations has been questioned, due to limitations such as the large influence of climate variations and the development of covered stadia (Ekstrand et al., 2006; FIFA, 2006; Veenbrink, 2002). Third generation artificial turf surfaces have been developed to address many of these limitations. The biomechanical evidence provided in chapter 3 demonstrated that when running on a third generation artificial surface, the peak rate of loading and pressure at the first metatarsal and peak pressure loading rate at the lateral heel were significantly reduced compared with a natural turf surface. Likewise, whilst turning, peak impact forces, were significantly lower on the artificial surface. These observations were made during May when the natural turf was described as hard and dry. As such, it was suggested that the third generation artificial surface may provide an alternative to natural turf when these surfaces are described as hard; conditions which have been associated with increased risk of injury during preseason.

Although participating on a third generation artificial playing surface has been shown to influence some biomechanical variables associated with overuse injury, the response of the soccer player may also be influenced by the construction of the playing surface. Third generation artificial turf surfaces are engineered with a firm base upon which a shock pad is placed beneath a top layer of artificial turf carpet, rubber and sand. This shock pad helps provide cushioning to the soccer player and can be made from varying materials and with different densities or thickness. McNitt et al. (2004) observed that when the shock pad is constructed with different materials and thickness, the mechanical cushioning can be significantly affected, although these differences can depend on the mechanical test being performed and the ratio of sand to rubber in-fill used (McNitt et al., 2004). However, despite mechanical cushioning tests providing a standardised and consistent result, they provide little information regarding how the participant will respond to the surface. Since human movement is complex and can be adapted to suit varying terrains (Bobbert et al., 1992; Denoth, 1986; De Wit & De Clercq, 1997; Dixon et al., 2005; Gerritsen et al., 1995), the response of an athlete to the surface condition is unlikely to correspond to the results obtained using mechanical tests.

The risk of overuse injury has also been linked to the type of footwear that is worn (Woods et al., 2002). Overuse Achilles tendon injury is a common injury in soccer (Woods et al., 2002). To lower the risk of sustaining a lower limb overuse injury, clinicians have used visco-elastic heel inserts to reduce and treat Achilles tendon pain and injury (Faunø et al., 1993; MacLellan & Vyvyan 1981; Nistor, 1981). However, despite the reported success of the heel insert intervention, little is known regarding the mechanism by which heel inserts work.

Along with heel inserts, cushioning insoles have been shown to reduce lower extremity pain and change biomechanical measurements associated with injury (Dixon et al., 2003; Gardener et al., 1988; House et al., 2002; Tooms et al., 1987). Sorbothane® have developed a cushioning insole for specific use inside a soccer boot. The manufacturer claims that these insoles are:

"Super slim shock absorbing insoles made from Sorbothane, an incredible visco-elastic polymer that soaks up heel strike and much of the vibration created every time your foot hits the ground. Amazingly, this not only takes the pressure off your feet, but it can significantly reduce much of the leg and back pain associated with hard exercise… and helps relieve those tired muscles and limbs." (Sorbopro, Appendix A).

These claims are not supported by published scientific research. Therefore, biomechanical research is needed to assess these claims and thus the potential for injury reduction.

One proposed mechanism by which the heel insert and insoles are successful in reducing injury risk is that with increased visco-elastic material placed into the shoe, significantly reduced force transients are experienced at impact (Light & MacLellan, 1971). MacLellan & Vyvyan, (1981) explained that the shock generated at impact travels through the lower extremity and is transferred from the bone to soft tissues such as the Achilles tendon. Therefore, the more visco-elastic the material at the heel, the greater the shock attenuation and the lower the internal force transient that transfers from bone to soft tissue. Similar mechanisms are suggested for the reduction of injury on more cushioned playing surfaces (Ekstrand & Gillquist, 1983; Ekstrand & Nigg, 1989).

To assess the influence of varying surface and footwear cushioning, peak impact forces, time of peak impact force, and peak rate of loading have been measured and have been able to distinguish between different levels of cushioning in soccer (Chapter 3; Dixon et al., 2003; Stiles & Dixon, 2006; Smith, Dyson & Janaway, 2004). Likewise, based on the findings of the previous chapter and past investigations (Hennig et al., 1996; Le Veau, 1992; Stiles & Dixon, 2007) there is particular focus on the effect of cushioning on the magnitude of peak pressures at the medial and lateral heel and the first and fifth metatarsal, under which the studs of soccer boots are positioned. Additionally, kinematic adaptations such as increased initial knee flexion at ground contact, (De Wit & De Clercq, 1997; Dixon et al., 2005; Wright et al., 1998), peak knee flexion and greater rearfoot movement (eversion when running and inversion during turning) (Kaelin et al., 1985; Nigg & Bahlsen, 1988; Stacoff et al., 1988) have been shown to occur on interfaces that have less cushioning

Another proposed method by which the heel insert is successful in the reduction of Achilles tendon pain and injury has been related to the insert causing the posterior portion of the foot to be raised relative to the forefoot (Clement et al., 1984; Leach et al., 1981) (Figures, 4.1 and 4.2). In providing this lift, the magnitude of peak dorsi-flexion during mid-stance may be lowered, indicating less tension of the triceps surae muscle group, particularly at the Achilles tendon (Dixon & Kerwin, 1998; 1999; 2002).

Figure 4.1. An example of plantar flexion without the use of a 10 mm heel insert.

Figure 4.2. An example of increased plantar flexion provided with the use of a 10 mm heel insert.

To assess such claims, Dixon and Kerwin (1999) taped a 7.5 mm high density EVA material to the plantar heel and measured the peak dorsi flexion and knee angle in participants when running, but found that these inserts did not influence the maximum dorsi flexion. In contrast, the 15 mm heel insert significantly reduced maximum dorsi flexion without altering peak knee flexion. The similar peak knee flexion angle indicated that the change in dorsi flexion was a consequence of the heel insert.

The similar excursions measured by Dixon and Kerwin (1999) with the smaller heel insert and no insert may have been influenced by the data being collected on barefooted participants. It is suggested that more realistic conditions may enable a significant reduction in peak dorsi-flexion which may explain the decreased injury/pain with the use of a heel insert intervention demonstrated in previous studies with smaller heel inserts (Faunø et al., 1993; MacLellan, 1985; MacLellan & Vyvyan, 1981).

In spite of the potential benefits of additional heel material, the risk of other injuries may be negatively influenced by this intervention. In particular, the introduction of a heel insert may influence the risk of metatarsal injury above the studs of the boots by accentuating the loading. The heel insert may also influence the risk of ankle inversion injury. In soccer, lateral ankle ligaments are most commonly damaged, with injury to the weak ATFL occurring in 66% to 73% of all ankle sprain cases (Woods et al., 2003; Sitler et al., 1994). An optimal level of inversion is necessary for impact attenuation

(Dayakidis & Boudolos 2006), yet the inclusion of the aforementioned devices may change the lateral stability of the ankle joint and allow the rearfoot to enter excessive inversion. As the inclusion of a heel insert may place the foot into a more plantar flexed position, this may also contribute to the loading of the ATFL. Further still, the magnitude of inversion and the subsequent loading can be accentuated by the increased plantar flexion as it allows the rearfoot to enter greater inversion magnitudes. Finally, the increased compliance on the lateral side of the heel insert may increase the susceptibility of the foot to enter into greater range of movement. This may also be problematic when cushioning insoles are used.

Turning movements have been described as complex tasks (Luethi et al., 1986). As such, measurements taken during the turn may not fully describe the biomechanical adaptations that occur in response to the surface and footwear conditions. In situations where the conditions are perceived as harmful, participants may make biomechanical changes prior to the turn, which may influence the kinematic and kinetic characteristics during the turn. As pressure insole technology allows multiple steps to be analysed, this technology can be used to understand any change in the plantar forces and pressure during the step made prior to the turn.

4.1.1 Aims and objectives

The aim of this research investigation is to compare two different shock pad densities to understand the biomechanical response of soccer players when running and turning under such conditions. The study presents a method by which artificial surfaces are tested and contributes towards the presently limited understanding of the effect of shock pad density of the playing surface on the performer. It is hypothesised that the measurements of lower extremity loading at impact will be significantly reduced on the more cushioned shock pad. This will be indicated by a reduced impact force (particularly during turning), time to peak impact force, peak rate of loading, and peak pressure and peak pressure loading rate at the medial and lateral heel during both running and turning. Forefoot loading will also be reduced on the more cushioned surface, measured by magnitude of the peak propulsive force, and peak pressure and peak pressure loading rate at the first and fifth metatarsals during both running and turning.

The study also aims to understand the biomechanical effect of placing a heel insert and cushioning insole into a soccer boot. Previously, both methods have been tested using biomechanical analysis, yet for the Sorbothane® insole designed specifically for use in soccer, no published biomechanical analysis is available to support their claims of shock attenuation. This study aims to contribute to supporting or rejecting these claims. Likewise, the mechanism behind heel insert success in reducing injury remains unclear. This has been associated with the limitations of previous investigations and so the present study aims to address these, in order to provide a clearer understanding of the injury protection mechanism. It is hypothesised that with the insert or insole placed inside the boot, there will be a significant reduction in the impact loading indicated by reduced impact force (particularly during turning), time to peak impact force, peak rate of loading and peak pressure and peak pressure loading rate at the medial and lateral heel (during both running and turning) compared to a control. Peak ankle dorsi-flexion is also expected to be significantly reduced when wearing the heel insert. There will also be a significant reduction of the forefoot loading (peak propulsive force, and peak pressure and peak pressure loading rate at the first and fifth metatarsals [during both running and turning]) when wearing the insole compared to a control and the heel insert.

Whilst the benefits of the heel insert may be evident, another aim of this study is to understand how this insert may influence the risk of lateral ankle and metatarsal injury. This potential for increased injury to other aspects of the lower extremity is often overlooked in research publications but is required, particularly if recommendations are to be provided regarding using these interventions to reduce injury in soccer. It is hypothesised that peak rearfoot inversion and plantar flexion will be significantly greater when turning with the heel insert placed inside the soccer boot, indicating an increased risk of lateral ankle damage. It is also hypothesised that greater forefoot pressures will be experienced with the heel insert, suggesting greater injury risk at the metatarsals.

4.2. Methods

4.2.1. Participants

Ten male participants (20.9 yrs [S.D. 2.5], 83.2 kg [S.D. 7.1], footwear size 10 -11) were asked to participate in this research investigation. The recruited participants were active local league soccer players. Each had recent experience of playing soccer on a third generation surface and was self-declared as being injury free for three months prior to the start of the testing. The participants were made aware of this exclusion criterion. They were told that they could drop out of the research project at any time, particularly if they were to feel pain during the task. All participants agreed to take part in the research investigation, understood the nature of the testing and the time commitment needed. This was confirmed by participants' signing an informed consent form. Ethics approval for the study was obtained from the ethics committee of the School of Sport and Health Sciences, University of Exeter.

In order to collect kinematic data, participants wore reflective markers that were placed at the hip, knee, lateral maleolus, fifth metatarsal, top of the foot and centre of the shin. Additionally, two markers were added to define the line of the calcaneous and two more to define the line of the shank, both of which were in the frontal plane and from behind. The placements of these markers were based on the recommendations of Soutas-Little et al. (1988) and are illustrated in Figures 4.3 and 4.4. These markers then allowed local reference planes to be constructed so that rotations about the joint could be accurately calculated regardless of their position in the global axis system (Appendix D).

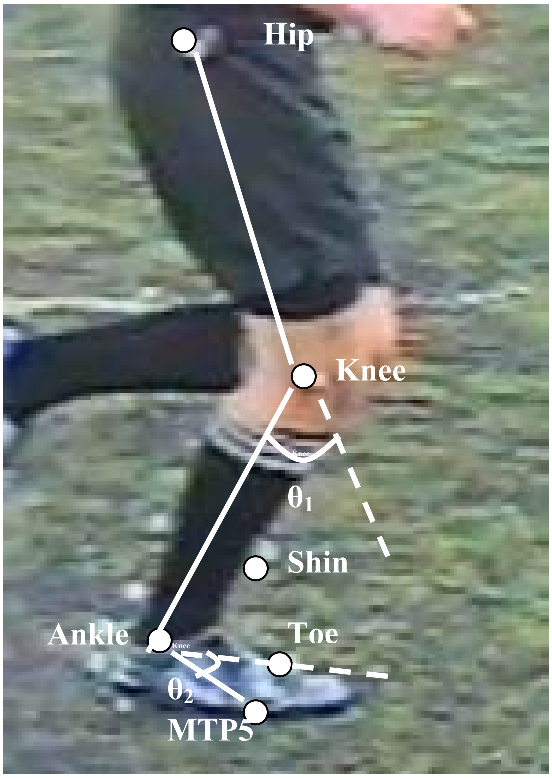


Figure 4.3. An example of the marker placements from the sagittal view along with a simplified indication of the anatomical angles for θ_1 (knee) and θ_2 (ankle)

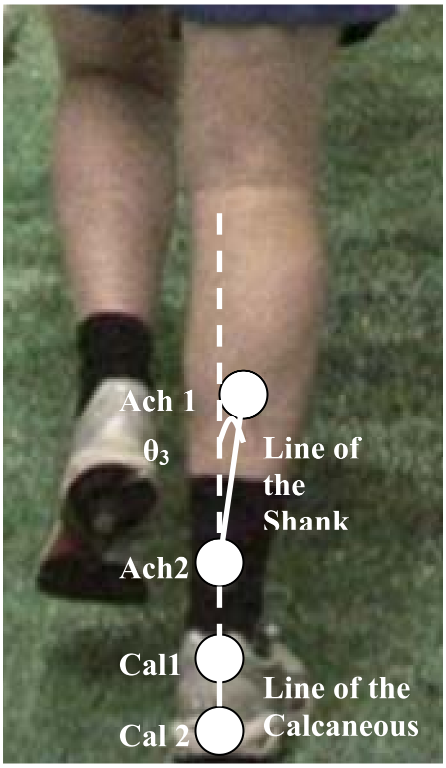


Figure 4.4. An example of the marker placements from the posterior view and the simplified calculated angles for θ_3 (rearfoot)

Two different shock pads were tested (Arpro® Expanded polypropylene BF2455W, 24mm S.D. 0.5mm thick, Brock International). These surfaces were reported by the manufacturer as having different cushioning ability (shock pad 1 = 55 g/litre, shock pad 2 = 65 g/litre). These surfaces were independently tested in a similar manner to that presented by Carré, James and Haake (2005). The procedure used a high speed camera (1000 Hz) and a mass of 2.1 kg dropped from a height of 15 cm (Figure 4.5). The results of this mechanical cushioning test supported those performed by the manufacturer, where higher peak deformations and lower peak force were experienced on the shock pad that the manufacturers had previously classified as the most cushioned (Table 4.1).

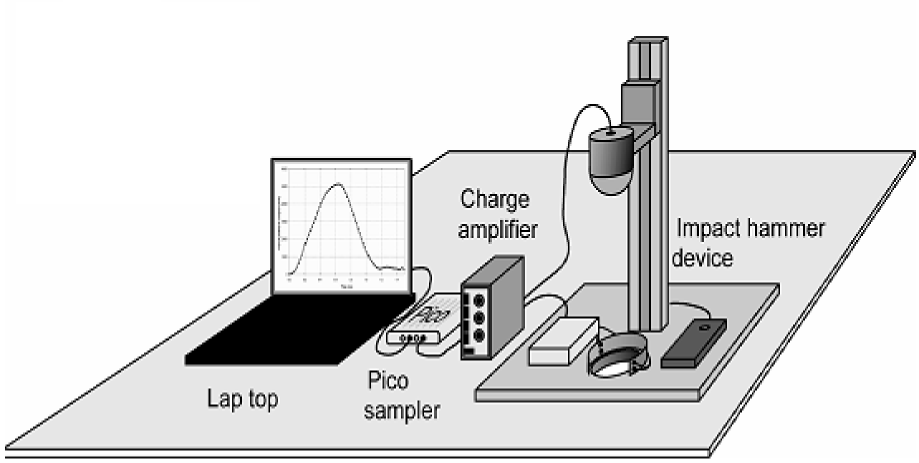


Figure 4.5. A diagram of mechanical drop test used to independently test the playing surface (adapted from Carré, et al., 2005)

Table 4.1 *Results of the mechanical surface test between the two shock pads. The table presents mechanical cushioning of the carpet alone, as well as the two shock pads both with the carpet place on top.*

	Artificial turf matting without a shock pad	Shock pad 1 55 g/litre	Shock pad 2 65 g/litre
Mean	1669.26 N	1238.5 N	1254.26 N
S.D.	56.34 N	46.09 N	48.45 N

Fifteen metres of each shock pad density was placed across a concrete floor in a biomechanics laboratory at the University of Exeter. Upon the shock pad, a length of third generation carpet was positioned which was of a similar length (Astroplay MXS 40, Lano sports, Herelbeke, Belgium). The same carpet was laid across both shock pad densities. Upon the carpet a 5:4 ratio of sand to rubber crumb was used (10 g/m^2 of sand and 8kg/m^2 of rubber crumb), as was recommended by the manufacturer. This was brushed to ensure uniformity across the carpet.

Participants were asked to perform test movements in three different footwear conditions. The first was a moulded soccer boot (Adidas, Copa Mundial) made up of 12 rubber/moulded studs designed for use on an artificial turf and which was used as a control condition (Figure 4.6). The second footwear condition consisted of the same soccer boots but with a Sorbothane cushioning insole placed inside of each boot (ProSole, Sorbo products division, Lancashire, UK) (Figure 4.7). The third footwear condition consisted of the same soccer boots but which had a 10 mm Sorbothane heel insert placed inside each boot (Sorbothane Shock Stopper heel pads, Sorbo products, Leyland, Lancashire, UK) (Figure 4.8). Each surface and footwear condition was tested in a random order for each participant.

Figure 4.6. An example of the Adidas Copa Mundial soccer boots worn by participants as the control.

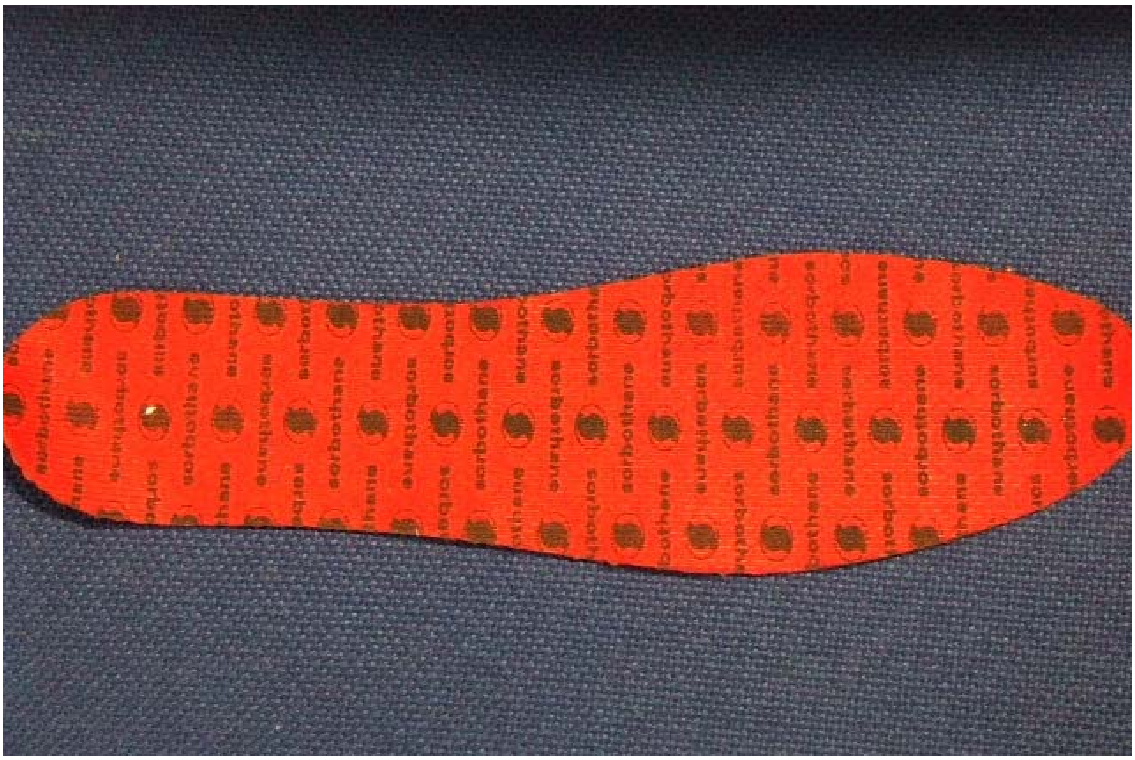


Figure 4.7. An example of the ProSole insole placed within the soccer boots worn by the participants

Figure 4.8. An example of the 10 mm heel insert placed within the soccer boots worn by the participants

4.2.2. Movements

Participants were asked to perform both running and turning movements. During the running task, participants were required to run the length of the test area (Figure 2.9) at a sub-maximal speed of 3.83 m/s (± 5%). The playing surface covered a force plate (AMTI, 960 Hz) within the ground. A square was marked on the surface to represent the location of the force plate and participants were required to place their right foot directly inside this square whilst maintaining their normal running gait. Also, during a turning movement, the participants were directed to run up to the marked box where they were asked to place their right foot, twist their hip and push off at the same speed and direction to which they came. No predetermined time was set for each participant during the turn. Instead, the speed of the first comfortable turn was then used for the participant's remaining trials. The speed of both the running and turning movements were monitored by placing photosensitive timing gates 1.5 metres either side of the force plate. Any running or turning trials that were not at the selected speed or consider being of a consistent style, were repeated.

Environmental set up

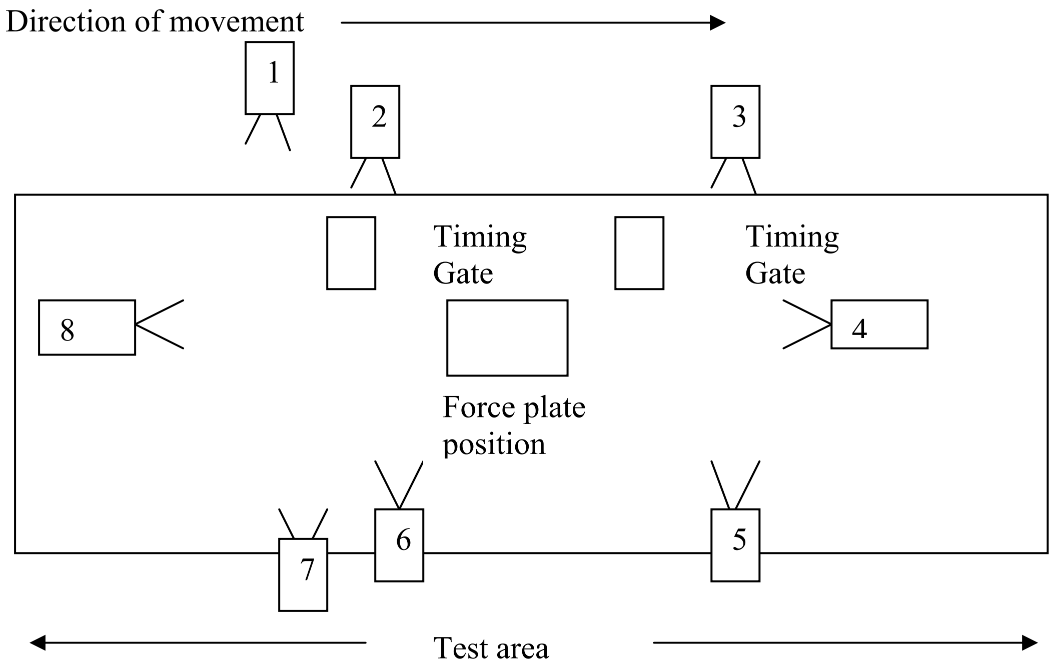


Figure 4.9. Schematic of the laboratory set up including the timing gates, cameras and force plate. Camera are represented by the boxes marked 1-8

Within the test environment, eight cameras (Pulnix, TM-6703 progressive scan, 120 Hz) (see Figure 4.9, 4.10 and 4.11) were focused on the test volume and synchronised with the force plate. This volume was calibrated by using a rigid object with markers of known locations and a wand with markers placed 0.38 cm apart, which was moved around within the calibration volume so that at least two cameras could see each marker at a given time. This provided the known and measured parameters which were later used in the DLT equation to reconstruct the three-dimensional coordinates of the markers worn by the participant.

Figure 4.10. An example of a Pulnix TM-6703 camera

Figure 4.11. A photographic view of laboratory set up

To collect in-shoe pressure data, a pair of Footscan pressure insoles (RSscan International, Belgium, 500 Hz) was inserted into the test footwear (Figure 3.7) which were used with the same methodology as presented in section 3.2.4.1.

4.2.3. Data analysis

The AMTI force plate (960 Hz) located beneath the sample of artificial turf provided the VGRF data that was used to calculate peak impact and propulsive forces, peak rate of loading (Equations 4.1 and 4.2) and time to peak vertical force during running and turning. The magnitudes of these variables were then used to compare the footwear and surface conditions.

$L_i = (x_i+1 - x_i-1) / 2\Delta t$ Equation 4.1

Where:

L_i= instantaneous loading rate

xi = vertical force for the ith field

ti = time

Δt = time interval

The footwear and surface variables were also compared by using the data from the pressure insoles. To quantify the pressure data, the Footscan insole software (version 6.345) was used and masks were set by the software to detect key areas of the foot (see Chapter 3, Figure 3.10). Peak pressure and peak pressure loading rate at the medial and lateral heel, and first and fifth metatarsals were calculated. Likewise, the pressure insole was used to collect peak impact and propulsive forces during running and turning. From the pressure insole, peak impact and propulsive forces and peak pressure and peak pressure loading rate were measured for the "braking step" (the step made prior to the turn).

After each trial had been collected, three dimensional co-ordinates were produced for each reflective marker worn by the participant. Three dimensional angle-time histories were then calculated for the knee, rearfoot and ankle which were smoothed using a quintic spline (Waltring, 1985). These angles were then referenced to the standing position and three-dimensional knee flexion angle at initial ground contact, peak knee flexion during mid-stance and peak dorsi-flexion were identified during both movements, along with peak rearfoot inversion and ankle plantar flexion during turning and rearfoot eversion during running (Figure 4.1 and 4.2).

4.2.4. Statistical analysis

During running the data for all 10 participants were used for statistical analysis, whereas during turning the data for only eight of the participants was used. This was because the data provided by two participants was not deemed to be of a high enough quality to be used for analysis. This was because not all of the markers were sufficiently tracked for the accurate analysis of the data. For each kinetic and kinematic parameters measured, an average of eight trials was used based on the stability of force plate data when running (Bates et al., 1983) and the reliability of pressure insole data whilst running and turning presented in Appendix B and C. There were two independent variables. One variable was footwear and the other was surface. Within each variable there were three footwear which included a control (moulded soccer boot), heel lift (moulded soccer boot with heel lift), and insole condition (moulded soccer boot with a cushioning insole) and two surface shock pad conditions with different densities. Therefore, a 3x2 repeated measures ANOVA was used to examine mean differences within each independent variable, as well as any interactions that occur between the independent variables. Mauchley's test of Sphericity was checked for each biomechanical parameter tested. When sphericity was violated (Mauchly's test $p<0.05$) the Greenhouse Geisser correction statistic was used within the main ANOVA test. A paired samples t-test with bonforroni correction was used as a post-hoc test to identify the location of any significant differences within the footwear variable ($p<0.05$). The alpha level was set at 0.05.

4.3. Results

4.3.1. Shock pad: Kinetic data

In this section the kinetic results measured during running are described. Figure 4.13 shows a typical vertical force-time history measured during running. The graph shows a double peaked shape, where the first peak represents the impact phase and the second indicates the propulsive action. Peak impact and propulsive forces collected with the insole were less than those measured with the force plate. The peak impact and propulsive force compared across the various footwear-surface conditions. It was shown that greater forces were observed whilst running on the less cushioned shock pad, although these differences were not significant. This trend was found whilst analysing data from both the force plate and insole. The peak rate of loading and the time of peak vertical forces were calculated using the data provided by the force plate. These variables were also not significantly different between the surface cushioning conditions (Table 4.2).

Peak pressure and peak pressure loading rate data at the first and fifth metatarsals and the medial and lateral heel were measured during running. Figure 4.14 shows the typical temporal and qualitative characteristics of the pressure-time history during ground contact at each of these sites. It is shown that the peak heel pressures occur first, although some pressure is placed on the first and fifth metatarsals during the early stance phase. The peak pressures at the first and fifth metatarsals occur during the latter stages of ground contact corresponding to the period of time when the participant is entering the propulsive phase. The results of the ANOVA RM comparing the two shock pad conditions for the measurements of peak pressure and the peak pressure loading rate at the medial and lateral heel and the first and fifth metatarsal are provided in Table 4.2. The results of this comparison showed that there were no significant differences between surface conditions.

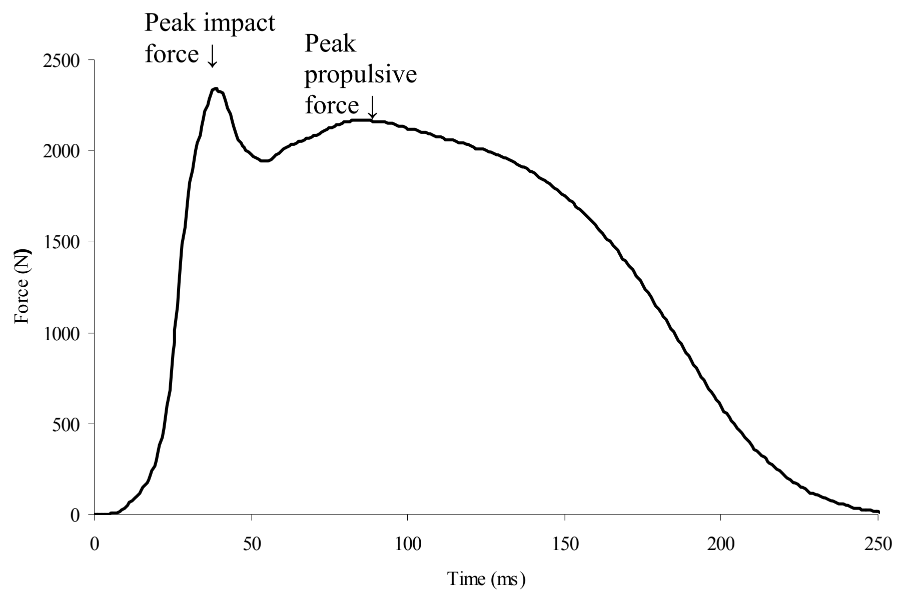


Figure 4.12. An example of the vertical force-time history from the force plate observed when running.

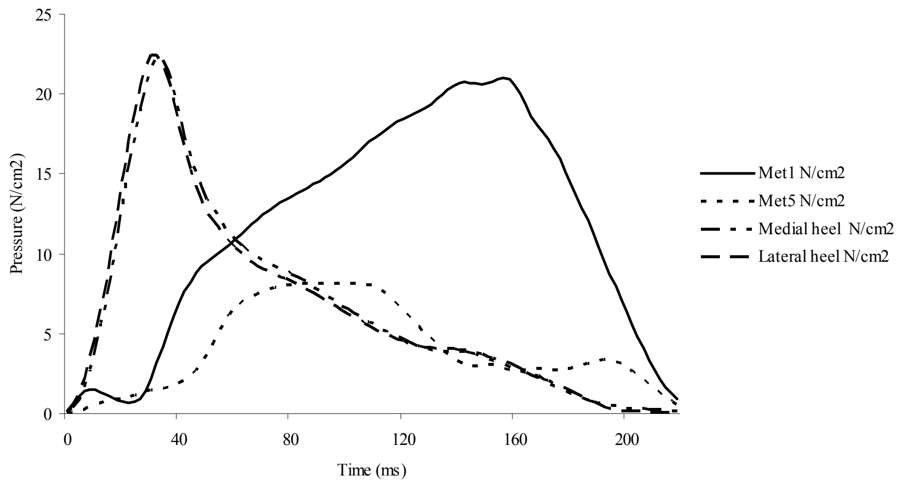


Figure 4.13. An example of the pressure-time history from the insole observed when running.

Table 4.2 *Kinetic data collected whilst running and turning and for the 'braking foot' on the two playing surfaces*

	Peak impact force (N) (force plate)	Peak impact force (N) (insole)	Time of peak impact force (ms)	Peak rate of loading (N/ms)	Peak Propulsive Force (N) (force plate)	Peak Propulsive Force (N) (insole)	Peak medial heel pressure (N/cm^2)	Peak lateral heel pressure (N/cm^2)	Peak fifth metatarsal pressure (N/cm^2)	Peak first metatarsal pressure (N/cm^2)	Peak medial heel loading rate (N/cm^2/ms)	Peak lateral heel loading rate (N/cm^2/ms)	Peak fifth metatarsal loading rate (N/cm^2/ms)	Peak first metatarsal loading rate (N/cm^2 /ms)
Run														
55g	1693.4	759.4	0.03	108656.0	2043.9	746.9	19.5	19.3	10.8	12.8	0.8	0.9	0.3	0.3
	(289.4)	(149.6)	(0.003)	(45120.6)	(225.05)	(99.8)	(5.6)	(7.5)	(6.1)	(3.9)	(0.4)	(0.3)	(0.2)	(0.1)
65g	1776.0	725.3	0.03	116442.9	2079.5	776.2	20.4	19.4	9.7	13.3	0.8	0.9	0.2	0.3
	(317.9)	(112.4)	(0.01)	(56511.1)	(226.74)	(83.4)	(5.4)	(4.5)	(3.1)	(5.3)	(0.2)	(0.2)	(0.1)	(0.1)
P	0.26	0.38	0.31	0.42	0.09	0.48	0.50	0.96	0.49	0.71	0.90	0.91	0.24	0.50
Turn														
55g	1619.7	749.6	0.04	145702.4	1221.0	494.7	37.2	16.0	5.00	14.7	2.0	0.8	0.15	0.5
	(271.8)	(188.0)	(0.01)	(30493.2)	(123.1)	(94.9)	(10.0)	(7.3)	(2.52)	(3.5)	(0.5)	(0.4)	(0.1)	(0.2)
65g	1587.4	824.1	0.04	138208.4	1198.9	486.9	37.4	16.6	6.3	22.3	1.9	0.8	0.20	0.7
	(249.4)	(107.5)	(0.01)	(59008.6)	(113.4)	(116.5)	(7.5)	(5.7)	(2.64)	(12.6)	(0.5)	(0.3)	(0.1)	(0.4)
	0.31	**0.05***	0.62	0.54	0.39	0.30	0.94	0.65	0.13	**0.01***	0.39	0.93	0.11	0.22
Braking														
55g	N/A	887.0	N/A	35.6	N/A	N/A	25.7	22.0	11.1	10.0	0.9	1.1	0.4	0.3
		(142.8)		(12.9)			(19.6)	(8.4)	(2.2)	(4.2)	(0.4)	(0.5)	(0.2)	(0.1)
65g	N/A	856.5	N/A	35.7	N/A	N/A	21.0	22.2	13.0	12.2	0.8	1.1	0.4	0.3
		(154.6)		(13.8)			(9.2)	(11.4)	(8.9)	(7.1)	(0.5)	(0.6)	(0.3)	(0.2)
P		0.49		0.98			0.28	0.95	0.25	0.19	0.81	0.93	1.00	0.12

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level.

Figure 4.15 shows a typical vertical force-time history measured by the force plate during turning. During the impact phase (the first 50 ms), the graph shows multiple peaks, the largest of these is classified as the peak impact force. During the mid-stance there is a lowering of the force and as the foot undergoes the propulsive phase there is a final peak shortly followed by the lessening of force back to zero.

Figure 4.16 indicates the time and shape of the pressure measurements at the medial and lateral heel and first and fifth metatarsals during turning. Peak heel pressures occurred at approximately the same time, although the magnitude of the medial heel pressure was greater than the lateral heel pressure. The shape of the metatarsal pressures time-history shows that the first metatarsal experiences a peak immediately following peak medial and lateral heel pressures. This is then reduced, but peaks again in the latter stages of ground contact as the participant enters the propulsive stages of gait. On the other hand, the pressure profile for the fifth metatarsal peaks at approximately the same time as the first metatarsal, but maintains a similar value until the propulsive phase.

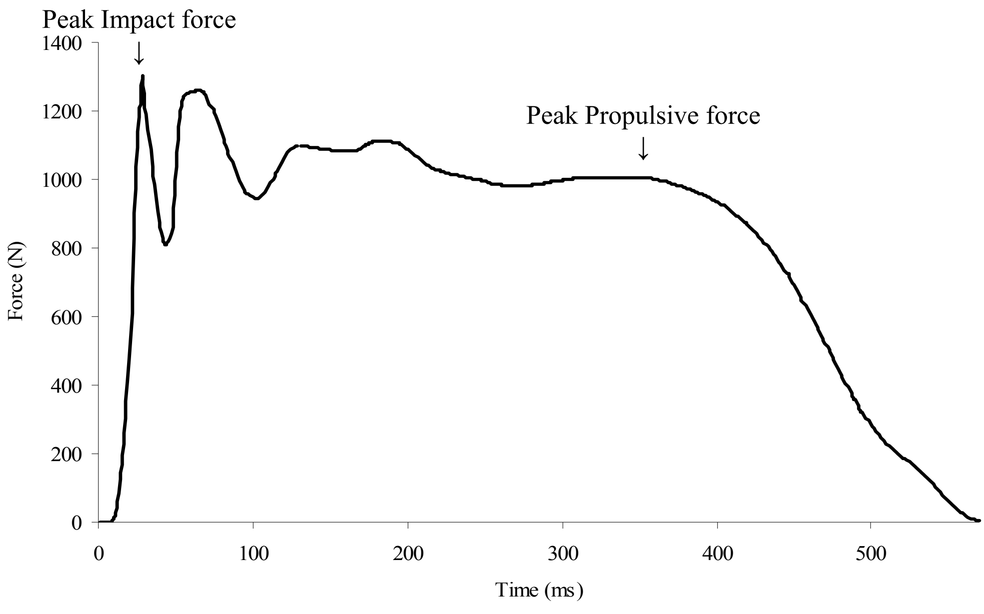


Figure 4.14. An example of vertical force-time history from the force plate occurring when turning.

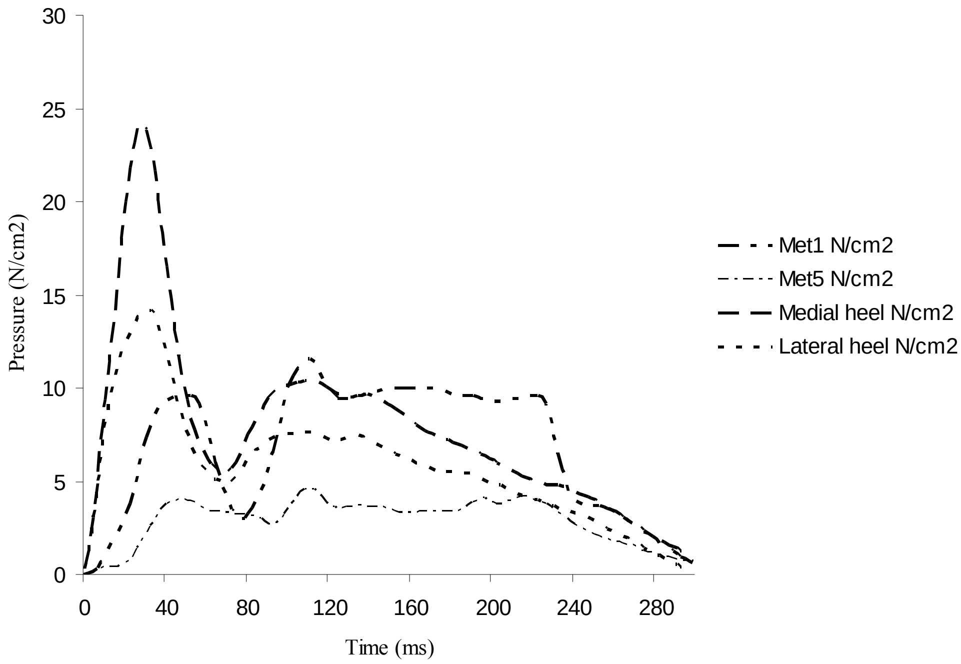


Figure 4.15. An example of the pressure-time history from the insole observed when turning.

Table 4.2 presents the group means for the measurement of peak impact forces by both the force plate and the insole. It was shown that the difference between the surfaces when turning was not significantly different when the measurement was collected with the force plate, whereas the peak impact force collected by the insole was significantly greater when turning on the 65g shock pad ($p = 0.05$). The mean time of peak impact force, peak rate of loading and peak propulsive force are also presented in Table 4.2. No significant differences were indicated between surface conditions for any of these measurements. The peak pressures and peak pressure loading rate at the first and fifth metatarsals and the medial and lateral heel are also shown in Table 4.2. The peak pressure at the first metatarsal was significantly greater when turning on the 65 g shock pad ($P = 0.01$).

The typical force-time history is presented in Figure 4.17 for the 'braking foot'. The participants experienced a sudden increase in force where it peaked and began to lower. This indicated the impact between the foot and the ground and the maximum magnitude of this force was used as the impact peak. A second peak was found later in the stance phase which represented the propulsive stage. Typical pressure time histories are presented in Figure 4.18 and show that the peak pressure at the fifth metatarsal and medial and lateral heel occurs approximately at the same time. Peak pressure at the first metatarsal occurs later in the stance phase and coincides with a second peak in the pressure profile of the fifth metatarsal. The magnitude of peak impact forces and peak rate of loading, as well as the peak pressures and peak pressure loading at the first and fifth metatarsals and medial and lateral heel, were measured from the pressure insole during this step. No significant differences were found for any of these variables under the different shock pad cushioning (Table 4.2, and 4.2).

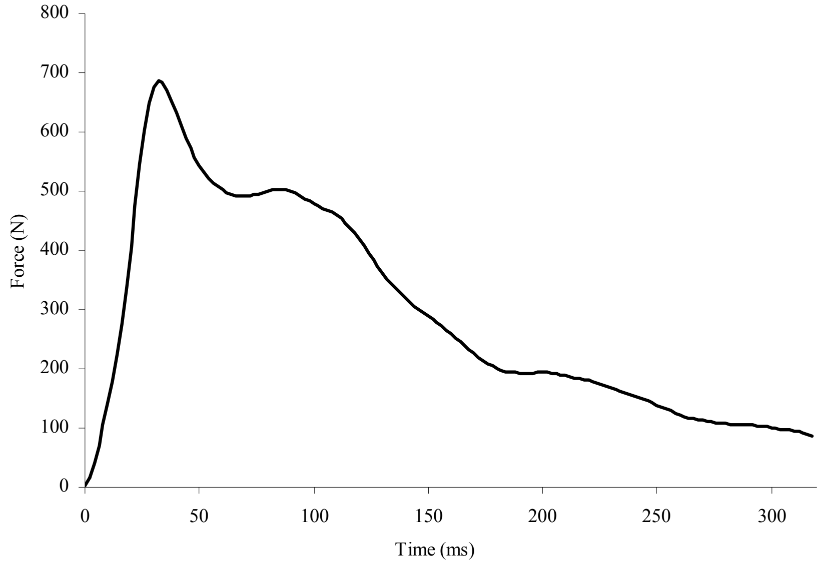


Figure 4.16. An example of vertical force-time history observed for the 'braking foot' during ground contact prior to turning.

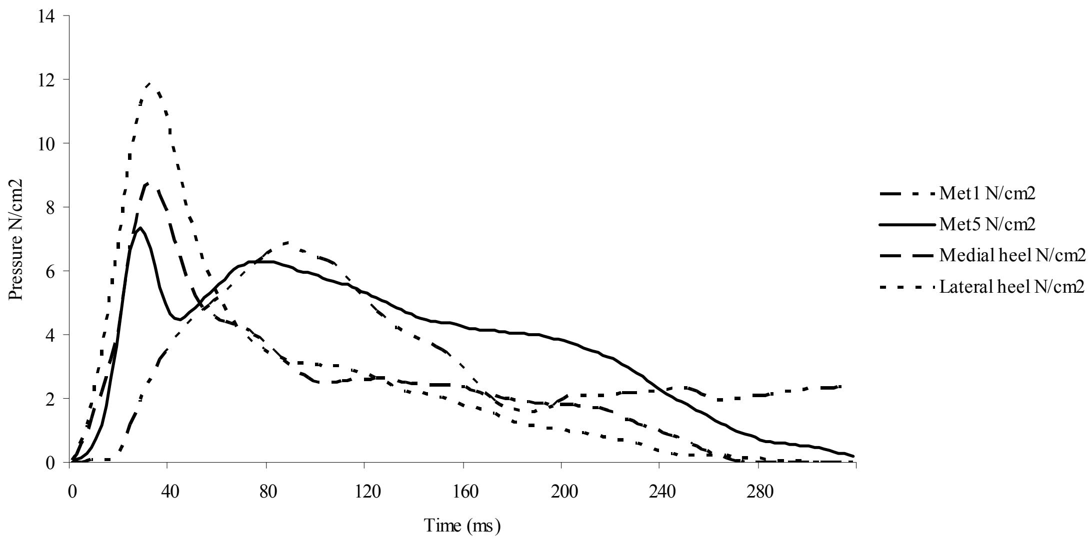


Figure 4.17. An example of the pressure-time history from the insole observed for the 'braking foot' during ground contact prior to turning.

4.3.2. Shock pad: Kinematic data

Figure 4.19 shows a typical angle-time history for the change in ankle angle occurring during running. It shows that immediately following contact the foot is plantar flexing until it reaches peak plantar flexion (lowest point of the initial portion of the graph), and then dorsi-flexes, as indicated by the upwards movement of the graph during mid-stance. Following peak dorsi-flexion (highest point), the foot again performs a plantar flexing movement up to toe-off. Figure 4.20 shows the angle-time history for the rearfoot. Ground contact was made where the rearfoot immediately everted, rolling into peak eversion position during mid-stance. During the propulsive stage the participants started to perform an inversion movement that caused the rearfoot to reduce the eversion angulations. Figure 4.21 shows the typical time history of the knee joint during running. The angle of the knee increases during mid-stance representing knee flexion prior to the propulsive phase where the knee extends.

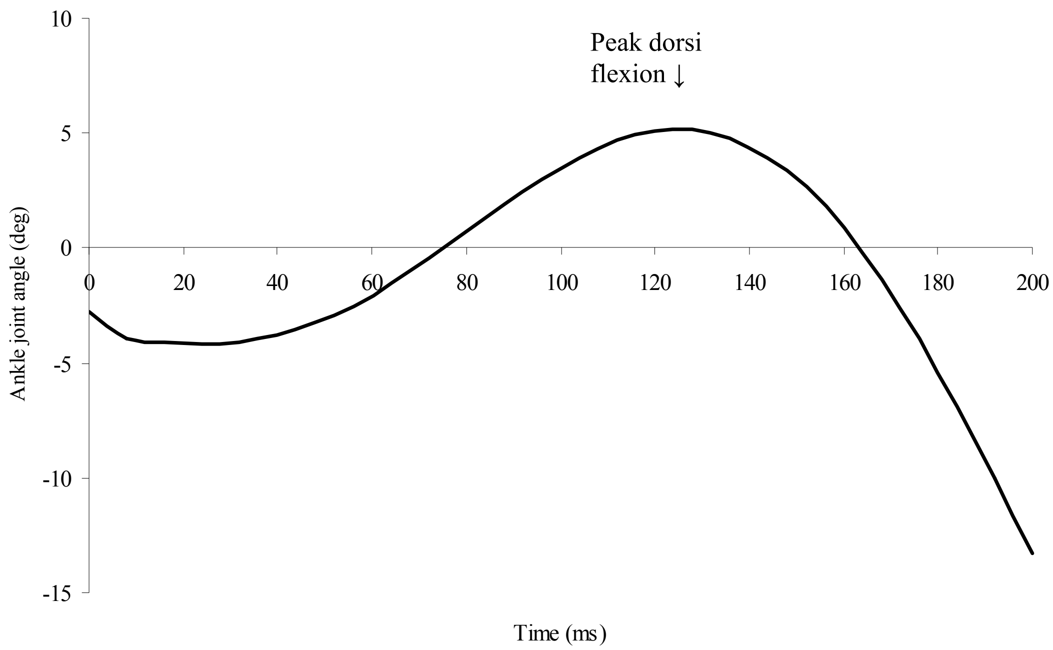


Figure 4.18. An example of ankle plantar flexion and dorsi-flexion time-history during ground contact when running.

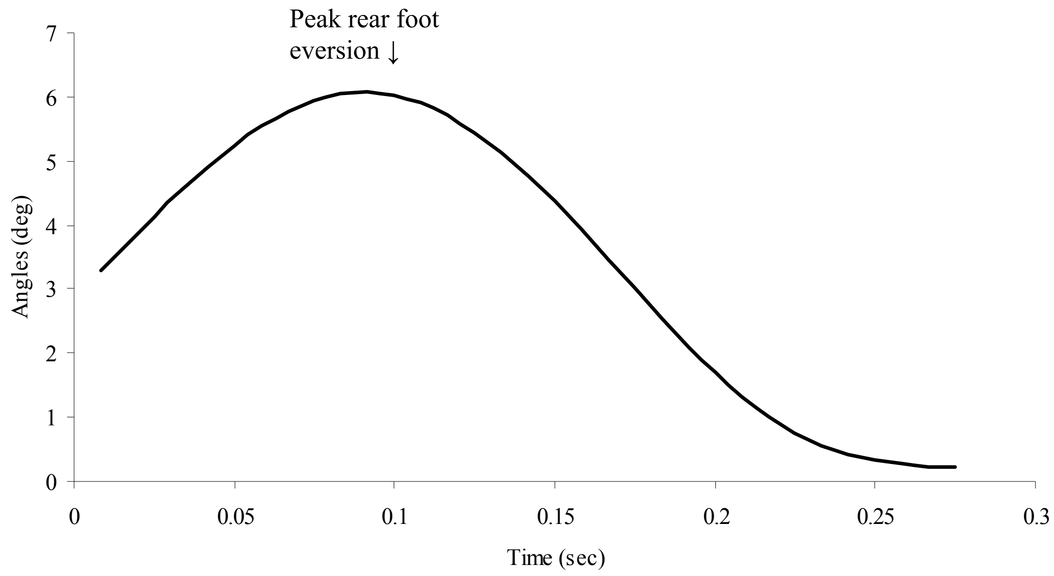


Figure 4.19. An example of ankle inversion/eversion time history during ground contact when running.

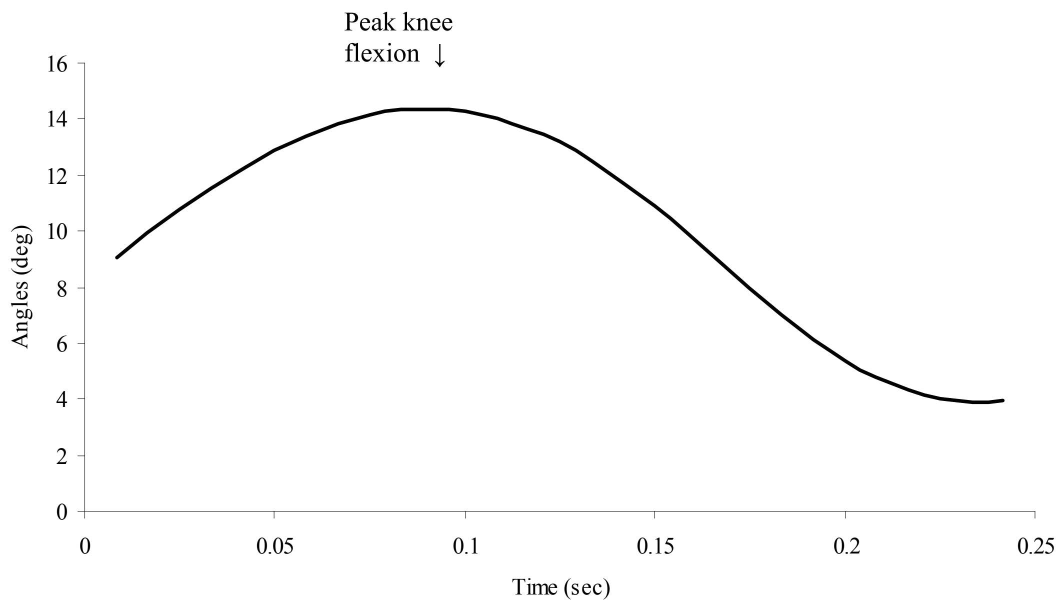


Figure 4.20. An example of knee flexion-extension time history during ground contact when running.

Peak dorsi-flexion, rearfoot eversion and knee flexion were compared between the different shock pad conditions and are presented in Table 4.3. The statistical analysis of the peak excursions found no significant kinematic differences between the surface conditions ($p < 0.05$).

Table 4.3 *Kinematic data collected whilst running and turning on the two shock pad conditions*

	Peak plantar flexion angle (deg)	Peak dorsi-flexion angle (deg)	Peak rearfoot angle (deg)	Knee Flexion at initial contact (deg)	Peak knee flexion angle (deg)
Run			Eversion		
55g	N/A	17. 4 (3.9)	4.7 (7.4)	6.7 (12.7)	24.0 (7.0)
65g	N/A	19.3 (4.9)	6.1 (6.2)	9.8 (10.7)	26.8 (7.5)
P		0.1	0.22	0.48	0.14
Turn			Inversion		
55g	-36.0 (12.4)	-13.5 (5.7)	-8.5 (6.1)	21.4 (11.8)	41.1 (15.8)
65g	-39.8 (14.1)	-13.9 (6.7)	-8.7 (7.0)	21.4 (18.0)	37.5 (15.5)
P	0.06	0.81	0.90	0.99	0.24

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

Figure 4.22 shows a typical rearfoot angle-time history during turning. The figure shows that during this movement, the participant lands and immediately begins an inversion movement. The foot then becomes less inverted during mid-stance but becomes more inverted as the foot moves through the propulsive phase. The angle-time history for the ankle joint is further presented in Figure 4.23. As the foot contacts the playing surface, the ankle's plantar flexion initially increases. The foot then dorsi flexes during mid-stance and plantar flexes during the propulsive phase. Figure 4.24 represents the time history of the knee angle during turning. Peak knee flexion occurs at approximately mid-stance and is represented by the greatest angle in the knee angle time history. Statistical differences between the shock pads were calculated on the kinematic data but found no differences for all measurements (Table 4.9).

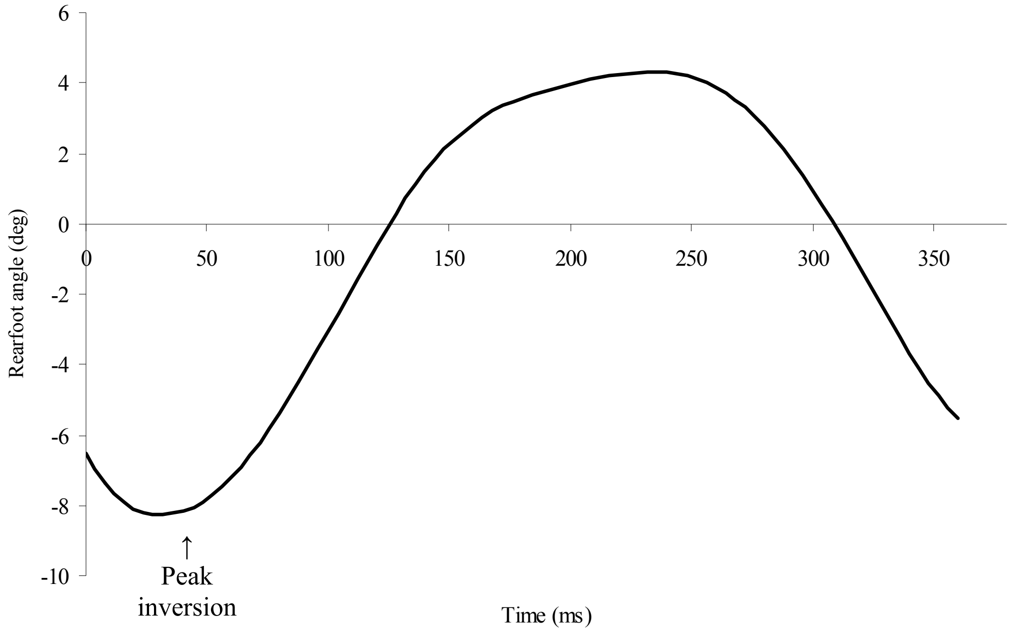


Figure 4.21. An example of a typical inversion/eversion movement-time history of during ground contact when turning.

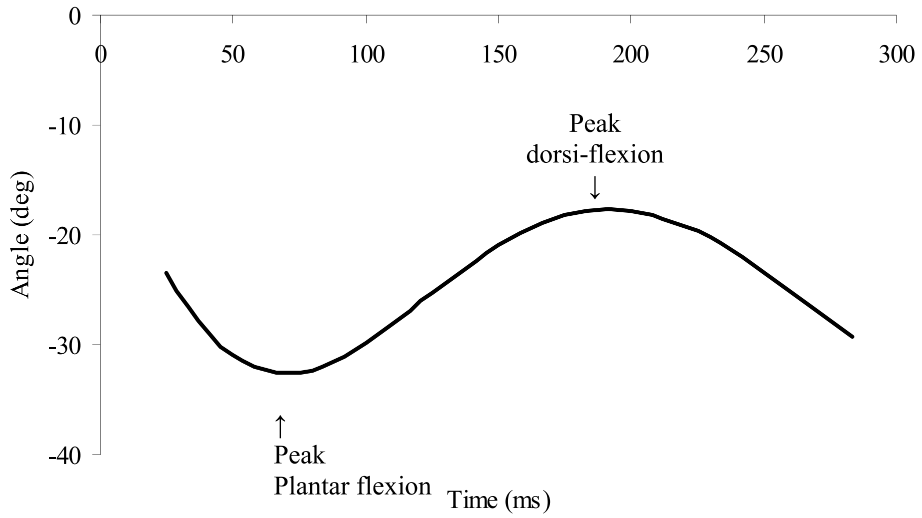


Figure 4.22. An example of a typical angle plantar flexion and dorsi-flexion-time history during ground contact when turning.

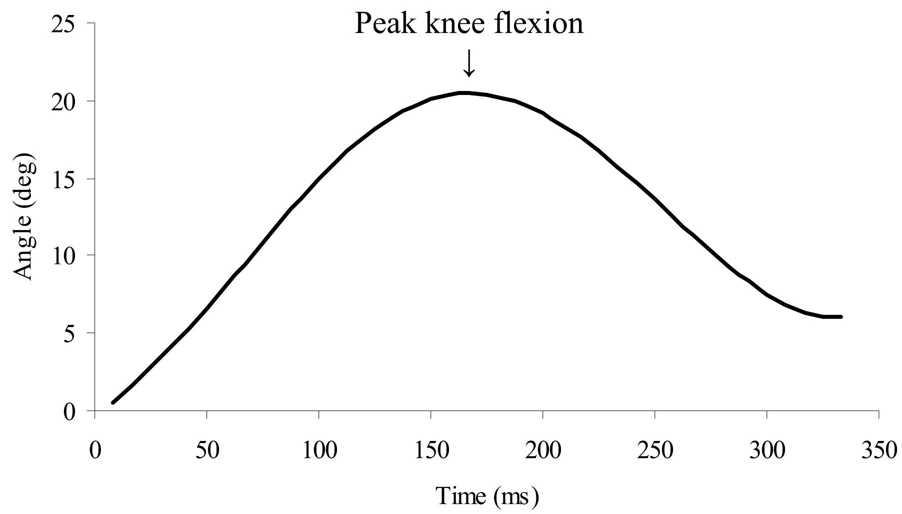


Figure 4.24. An example of a typical knee flexion-extension time history during ground contact when turning.

4.3.3. Footwear: Kinetic data

Peak impact force and peak propulsive forces (from the insole and force plate), and the peak rate of loading and time of peak impact force were also compared when running and turning in the different footwear conditions (Table 4.4). The time to peak impact force was significantly greater in the insert condition compared to the control (mould soccer boot) and insole condition when tuning ($p = 0.03$), although no other significant differences could be detected between any of the footwear conditions when running or turning. Table 4.11 shows the peak pressures and peak pressure loading rates occurring at the first and fifth metatarsal joint and the medial and lateral heel when running and turning. There were no significant differences between footwear conditions for any of these variables ($P > 0.05$). Peak impact force and peak rate of loading, as well as peak pressure and peak pressure loading rate collected from the pressure insole during the 'braking step' are also presented in Table 4.4. For each of these measurements, there were no significant differences.

Table 4.4 *Kinetic data collected whilst running and turning and for the 'braking foot' in the control, heel insert and cushioning insole conditions*

	Peak impact force (N) (force plate)	Peak impact force (N) (insole)	Time of peak impact force (ms)	Peak rate of loading (N/ms)	Peak Propulsive Force (N) (force plate)	Peak Propulsive Force (N) (insole)	Peak medial heel pressure (N/cm^2)	Peak lateral heel pressure (N/cm^2)	Peak fifth metatarsal pressure (N/cm^2)	Peak first metatarsal pressure (N/cm^2)	Peak medial heel loading rate (N/cm^2/ms)	Peak lateral heel loading rate (N/cm^2/ms)	Peak fifth metatarsal loading rate (N/cm^2/ms)	Peak first metatarsal loading rate (N/cm^2/ms)
Run														
	1704.3	744.5	0.03	116793.2	2029.6	766.9	19.9	20.3	10.9	13.5	0.9	0.9	0.18	0.48
Control	(354.6)	(144.3)	(0.004)	(66203.6)	(208.8)	(94.5)	(4.7)	(6.9)	(6.7)	(4.5)	(0.5)	(0.3)	(0.09)	(0.28)
	1728.3	743.1	0.03	105800.1	2082.6	761.5	19.4	19.2	10.4	12.6	0.8	0.9	0.17	0.62
Heel insert	(270.6)	(137.7)	(0.004)	(41006.0)	(240.5)	(88.2)	(6.0)	(6.4)	(4.5)	(4.9)	(0.3)	(0.3)	(0.24)	(0.39)
Cushioning	1771.6	739.5	0.03	115054.9	2072.7	756.2	20.5	18.6	9.4	13.0	0.8	0.9	0.18	0.61
insole	(293.3)	(120.7)	(0.006)	(43492.7)	(231.7)	(98.5)	(5.9)	(5.2)	(2.3)	(4.4)	(0.2)	(0.2)	(0.16)	(0.30)
P	0.36	0.97	0.32	0.46	0.1	0.60	0.67	0.22	0.36	0.41	0.80	0.50	0.18	0.39
Turn														
	1556.7	791.9	0.039	149149.8	1203.3	500.4	40.2	16.6	5.5	19.3	2.1	0.8	0.2	0.62
Control	(231.4)	(137.7)	(0.03)	(64180.8)	(123.1)	(85.9)	(8.2)	(7.1)	(2.9)	(11.6)	(0.4)	(0.4)	(0.1)	(0.4)
	1563.9	781.8	0.042	132712.2	1229.2	485.9	36.2	15.5	5.6	17.7	1.8	0.8	0.2	0.61
Heel insert	(246.4)	(124.8)	(0.04)	(32777.0)	(121.4)	(122.4)	(8.0)	(5.5)	(2.4)	(10.5)	(0.4)	(0.3)	(0.2)	(0.3)
Cushioning	1641.7	777.5	0.039	144004.2	1197.3	487.0	35.5	16.8	5.7	18.4	2.0	0.8	0.2	0.48
insole	(275.3)	(150.0)	(0.03)	(38280.9)	(113.4)	(110.2)	(9.8)	(7.1)	(3.0)	(8.0)	(0.6)	(0.4)	(0.1)	(0.3)
	0.13	0.42	**0.03***	0.46	0.37	0.88	0.11	0.46	0.87	0.71	0.33	0.87	0.91	0.34
Braking														
		865.7		34.7			22.9	22.0	12.2	11.5	0.9	1.1	0.5	0.3
Control	N/A	(136.5)	N/A	(13.4)	N/A	N/A	(11.0)	(9.0)	(3.1)	(5.8)	(0.5)	(0.6)	(0.2)	(0.1)
		875.3		35.2			20.7	22.7	13.3	10.3	0.8	1.0	0.4	0.3
Heel insert	N/A	(151.1)	N/A	(12.5)	N/A	N/A	(8.8)	(12.1)	(10.5)	(6.5)	(0.3)	(0.4)	(0.2)	(0.2)
Cushioning		875.4		37.0			26.6	21.6	10.6	11.4	0.9	1.1	0.4	0.3
insole	N/A	(163.9)	N/A	(14.3)	N/A	N/A	(22.8)	(8.9)	(3.0)	(5.5)	(0.4)	(0.5)	(0.2)	(0.1)
P		0.98		0.87			0.54	0.95	0.41	0.83	0.63	0.65	0.31	0.26

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

4.3.4. Footwear: Kinematic data

The mean kinematic data collected during running and turning were compared between the footwear conditions and are presented in Table 4.5. No significant differences were observable for any of the measurements when running, although plantar flexion was significantly greater for the heel lift condition compared to the control and insole conditions when turning (p = 0.01) (Table 4.5).

Table 4.5 *Kinematic data collected whilst running and turning in the control, heel insert and cushioning insole conditions*

	Peak plantar flexion angle (deg)	Peak dorsi-flexion angle (deg)	Peak rearfoot angle (deg)	Knee flexion at initial contact (deg)	Peak knee flexion angle (deg)
Run			Eversion		
Control	N/A	18.2 (4.8)	4.5 (6.2)	6.1 (13.7)	24. 0 (8.2)
Heel lift	N/A	18.5 (4.2)	6.5 (7.7)	8.5 (12.1)	25.5 (6.3)
Insole	N/A	18.4 (4.7)	5.4 (6.7)	10.2 (9.4)	26.6 (7.5)
P		0.92	0.37	0.18	0.27
Turn			Inversion		
Control	-38.9 (12.0)	-14.8 (5.3)	-7.3 (6.9)	25.3 (16.2)	37.9 (16.9)
Heel lift	-43.1 (14.0)	-12.1 (5.8)	-10.3 (7.2)	20.8 (14.3)	41.4 (14.5)
Insole	-34.1 (13.0)	-14.3 (7.2)	-8.2 (7.0)	18.3 (14.2)	38.6 (15.9)
P	**0.01***	0.25	0.93	0.06	0.07

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the p<0.05 level

4.3.5. Footwear-surface interaction: kinetic data

Table 4.6 presents the interaction between the footwear and surface conditions for the kinetic data when running. There was a significant interaction presented between the footwear and surface variables when running for the measurement of peak impact forces (P = 0.04), however, post hoc test revealed that the differences were not significant. There were no other significant interactions when either running or turning or for the braking foot.

Table 4.6 *Kinetic data interactions between the footwear (control, heel insert and cushioning insole conditions) and surface (55g and 65g) whilst running and turning and for the braking foot.*

		Peak impact force (N) (force plate)	Peak impact force (N) (insole)	Time of peak impact force (ms)	Peak rate of loading (N/ms)	Peak Propulsive Force (N) (force plate)	Peak Propulsive Force (N) (insole)	Peak medial heel pressure (N/cm^2)	Peak lateral heel pressure (N/cm^2)	Peak fifth metatarsal pressure (N/cm^2)	Peak first metatarsal pressure (N/cm^2)	Peak medial heel loading rate (N/cm^2/ms)	Peak lateral heel loading rate (N/cm^2/ms)	Peak fifth metatarsal loading rate (N/cm^2/ms)	Peak first metatarsal loading rate (N/cm^2/ms)
Run	**Run**														
Control	55g	1688.67	776.47	0.031	108657.69	2035.02	744.48	19.52	20.92	12.12	13.29	0.95	0.92	0.15	0.39
		(328.41)	(162.58)	(0.003)	(48941.71)	(218.39)	(89.05)	(5.37)	(9.06)	(8.550	(4.82)	(0.70)	(0.36)	(0.09)	(0.24)
Heel insert	55g	1638.14	747.95	0.032	100055.17	2048.99	753.41	17.95	18.16	10.79	12.33	0.68	0.79	0.13	0.51
		(247.76)	(164.03)	(0.003)	(39982.30)	(239.96)	(97.29)	(5.37)	(7.45)	(5.96)	(3.44)	(0.24)	(0.31)	(0.1)	(0.54)
Cushioning insole	55g	1753.43	753.82	0.031	117254.99	2047.53	742.68	20.94	18.84	9.57	12.85	0.84	0.86	0.17	0.55
		(305.23)	(135.21)	(0.004)	(49008.15)	(240.40)	(120.88)	(6.09)	(6.30)	(2.69)	(3.54)	(0.19)	(0.27)	(0.18)	(0.20)
Control	65g	1719.93	712.58	0.031	124928.72	2035.02	789.29	20.24	19.70	9.74	13.75	0.78	0.86	0.22	0.58
		(396.35)	(123.55)	(0.005)	(81917.38)	(218.39)	(99.12)	(4.16)	(4.14)	(4.38)	(4.47)	(0.18)	(0.18)	(0.1)	(0.39)
Heel insert	65g	1818.44	738.30	0.031	111545.10	2048.99	769.66	20.95	20.22	9.92	12.89	0.90	0.91	0.20	0.74
		(274.13)	(114.45)	(0.004)	(43334.67)	(239.96)	(82.54)	(6.42)	(5.31)	(2.65)	(6.30)	(0.28)	(0.23)	(0.14)	(0.17)
Cushioning insole	65g	1789.81	725.15	0.035	112854.81	2047.53	769.76	20.06	18.35	9.31	13.16	0.85	0.84	0.19	0.67
		(296.17)	(109.54)	(0.01)	(39759.99)	(240.40)	(73.94)	(5.96)	(4.13)	(1.92)	(5.35)	(0.27)	(0.20)	(0.14)	(0.38)
P		**0.04***	0.23	0.13	0.29	0.36	0.65	0.28	0.36	0.37	0.99	0.18	0.17	0.41	0.17
Turn															
Control	55g	1628.07	743.95	0.043	146078.36	1193.23	530.8	38.97	17.23	4.76	15.29	2.04	0.85	0.13	0.51
		(223.25)	(150.27)	(0.014)	(28307.39)	(94.28)	(96.5)	(6.70)	(8.25)	(2.86)	(3.58)	(0.25)	(0.43)	(0.11)	(0.24)
Heel insert	55g	1557.37	736.57	0.034	136747.40	1271.67	470.0	37.83	15.14	5.64	12. 78	1.94	0.75	0.17	0.55
		(279.22)	(123.90)	(0.01)	(27596.71)	(131.54)	(66.3)	(10.39)	(6.22)	(2.17)	(3.24)	(0.53)	(0.35)	(0.18)	(0.20)
Cushioning	55g	1673.76	749.59	0.036	154281.28	1198.08	480.5	34.82	15.47	4.46	15.93	2.10	0.78	0.15	0.39

insole		(327.10)	(188.020	(0.009)	(36281.77)	(138.61)	(84.5)	(12.81)	(8.13)	(2.65)	(3.34)	(0.73)	(0.44)	(0.09)	(0.17)
Control	65g	1485.24	839.83	0.041	152221.19	1213.32	489.4	41.45	15.92	6.21	23.27	2.09	0.78	0.20	0.74
		(230.84)	(113.13)	(0.009)	(89464.71)	(152.81)	(157.8)	(9.69)	(6.29)	(2.85)	(15.41)	(0. 53)	(0.37)	(0.14)	(0.54)
Heel insert	65g	1570.46	827.03	0.04	128676.98	1186.78	531.3	34.61	15.94	5.58	22.60	1.74	0.74	0.19	0.67
		(228.12)	(115.60)	(0.01)	(38773.03)	(100.78)	(114.1)	(4.79)	(5.02)	(2.25)	(13.01)	(0. 22)	(0.29)	(0.14)	(0.39)
Cushioning	65g	1609.67	805.38	0.038	133727.05	1196.46	442.7	36.27	18.05	6.98	20.93	1.86	0.87	0.22	0.58
insole		(230.44)	(105.39)	(0.01)	(39779.71)	(91.26)	(92.4)	(6.32)	(6.29)	(2.92)	(10.63)	(0..50)	(0.38)	(0.10)	(0.38)
P		0.09	0.93	0.26	0.58	0.32	0.29	0.59	0.22	0.18	0.51	0.57	0.50	0.62	0.86
Braking															
Control	55g	N/A	888.67		35.48			24.97	22.64	12.36	10.74	0.85	1.08	0.47	0.30
			(28.22)	N/A	(13.99)	N/A	N/A	(11.35)	(8.16)	(2.50)	(4.58)	(0.30)	(0.45)	(0.17)	(0.13)
Heel insert	55g	N/A	867.73		34.57			21.32	21.61	10.36	8.13	0.80	0.99	0.36	0.20
			(165.06)	N/A	(12.72)	N/A	N/A	(9.52)	(8.91)	(1.92)	(2.45)	(0.38)	(0.46)	(0.13)	(0.07)
Cushioning	55g	N/A	904.73		36.82			30.90	21.89	10.59	11.21	0.91	1.10	0.40	0.27
insole			(147.53)	N/A	(13.52)	N/A	N/A	(31.21)	(9.15)	(1.92)	(4.91)	(0.41)	(0.50)	(0.16)	(0.10)
Control	65g	N/A	842.74		33.84			20.75	21.43	12.11	12.33	0.87	1.15	0.48	0.36
			(148.15)	N/A	(13.5)	N/A	N/A	(10.84)	(10.22)	(3.82)	(6.97)	(0.61)	(0.70)	(0.24)	(0.16)
Heel insert	65g	N/A	883.70		35.99			19.98	23.86	16.68	12.68	0.72	0.94	0.38	0.32
			(144.43)	N/A	(13.10)	N/A	N/A	(8.4)	(15.51)	(14.94)	(8.71)	(0.30)	(0.44)	(0.29)	(0.19)
Cushioning	65g	N/A	846.05		37.27			22.22	21.34	10.66	11.60	0.88	1.12	0.38	0.27
insole			(182.74)	N/A	(15.94)	N/A	N/A	(9.13)	(9.23)	(3.92)	(6.41)	(0.45)	(0.55)	(0.26)	(0.15)
P			0.75		0.95			0.79	0.87	0.24	0.57	0.95	0.94	0.95	0.46

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability and * denotes a significant difference at the $p<0.05$ level

4.3.6. Footwear-surface interaction: kinematic data

Table 4.7 presents the interactions between the footwear and surface conditions for the kinematic variables taken. However, there were no significant differences for any of the footwear and surface interactions ($p > 0.05$).

Table 4.7 *Kinematic data interactions between the footwear (control, heel insert and cushioning insole conditions) and surface (55g and 65g) whilst running and turning*

		Peak plantar flexion angle (deg)	Peak dorsi-flexion angle (deg)	Peak rearfoot angle (deg)	Knee flexion at initial contact (deg)	Peak knee flexion angle (deg)
Run				Eversion		
55g	Control	N/A	17.6 (3.7)	7.2 (9.8)	3.1 (15.7)	24.7 (6.5)
55g	Heel lift	N/A	19.3 (4.7)	5.8 (5.1)	7.2 (14.5)	26.3 (6.4)
55g	Insole	N/A	16.6 (3.4)	4.3 (5.5)	9.8 (8.0)	25.6 (5.9)
65g	Control	N/A	20.2 (5.2)	6.4 (7.9)	9 (11.7)	27.5 (9.1)
65g	Heel lift	N/A	17.9 (4.8)	2.7 (6.4)	9.8 (9.6)	21.5 (8.4)
65g	Insole	N/A	18.5 (5.1)	6.2 (5.9)	10.6 (10.8)	26.6 (7.7)
	P		0.16	0.27	0.64	0.53
Turn				Inversion		
55g	Control	-39.2 (5.7)	-12.8 (5.9)	-11.3 (6.4)	25.7 (13.8)	42.5 (13.8)
55g	Heel lift	-47.0 (11.8)	-11.4 (5.9)	-9.4 (8.2)	17.8 (12.4)	40.3 (16.0)
55g	Insole	-30.8 (10.2)	-13.3 (6.3)	-7.3 (6.3)	20.8 (9.2)	39.1 (19.7)
65g	Control	-37.4 (15.3)	-15.3 (8.2)	-9.2 (7.9)	24.8 (18.5)	38.1 (12.5)
65g	Heel lift	-38.1 (10.4)	-14.5 (5.4)	-7.0 (5.4)	23.8 (16.2)	41. 7 (15.2)
65g	Insole	-39.8 (14.1)	-15.0 (5.7)	-7.6 (8.5)	15.7 (19.2)	34.2 (18.7)
	P	0.75	0.19	0.06	0.36	0.35

Data are reported as mean and standard deviation (in brackets). Where, P denotes the statistical probability.

4.4. Discussion

This study provides a novel investigation into the effect of different shock pad densities and footwear cushioning on the biomechanical measurements associated with acute and overuse injury in football. It was observed that significant differences were present between the independent variables. Suggested reasons for and implications of these changes are now discussed.

4.4.1. Playing Surface

Variation in the playing surface shock pad has been described in past literature as effective in changing the mechanical cushioning provided to the athlete (McNitt et al. 2004). The observation that peak impact forces when running were not significantly different between the shock pads in the current study, fails to support such evidence, yet this finding is consistent with past research investigating the biomechanical response to changes in surface cushioning (Dixon & Stiles, 2003; Dixon et al., 2008).

Impact forces are determined from the resultant vertical force vector, which represents the change in acceleration of the body's COM. The contribution of the 'rest of the body' to this acceleration is larger than the contribution of the support leg (Bobbert et al., 1992). This may cause the peak VGRF to be insensitive to a change in acceleration of the lower extremity and thus fail to show differences between surfaces of different mechanical cushioning. Likewise, the summation of both active and passive forces during the impact phase may mask the ability of mechanical surface cushioning to reduce the passive forces, which load the lower extremity to cause injury (Hamill, 1996). The rate of change in force application has been shown to indicate the cushioning provided by the surface and has been linked to the risk of injury (Dixon et al., 2003; Lees & McCullagh, 1984; Nigg et al., 1987; Dixon & Stiles, 2003), yet these measurements were also not significantly different in the current study when running.

For a heel-strike ground contact, most force generated at impact is at the heel. However, as previously highlighted, the resultant impact force is the summation of heel as well as some mid- and forefoot force (Shorten, 2002). Consequently, localised heel forces or pressures may offer greater insight and be more sensitive to surface changes than measures of resultant forces, although no direct causal relationship between the pressure patterns experienced by participants and specific overuse injuries has been established (Eils et al., 2004). Despite this, isolation of heel forces and pressure have been proven a more sensitive method for comparing shoe and surface conditions (natural and third generation artificial turf) in comparison to measurements using resultant GRF (Dixon et al., 2008; Ford et al., 2008; Chapter 3). Although this may have been true in these previous studies, the comparisons of plantar pressures at the medial and lateral heel were also not significantly different in the current study during running. One alternative hypothesis to explain these results could have been that in response to differently cushioned surfaces, participants made kinematic adaptations in order to maintain similar forces (Ferris et al., 1998; Gerritsen et al., 1995). However, no kinematic differences were identified either.

In light of this accumulated data, the evidence suggests that the surfaces may not have been distinct enough to elicit differences for any of these study measurements during running. However, it is possible that the role of the shock pad will become more important as the surface ages. Mechanical tests have shown that the condition of the infill material primarily influences the force reduction, vertical deformation and ball bounce, which have all been related to impact cushioning (Alcantara et al., 2004). Alcantara et al. (2004) found that wear by use (number of cycles) changes the mechanical properties of infill resulting from the compaction of the rubber. This occurred with repeated use of the artificial surface over a long period of time and can significantly reduce the values of all parameters that define deformation and force reduction (Alcantara et al., 2006). This wear has been shown to be caused by damage to artificial surfaces over time (Sifers & Beard, 1994). As a consequence, third generation soccer pitches with long carpet pile filled with rubber are not considered to be greatly effected by the shock pad until infill material compaction has occurred, which typically takes several years (Fleming et al., 2008). McNitt et al., (2004) showed that the combination of sand and rubber determines whether any significant differences in mechanical parameters of cushioning were shown between two shock pad conditions. When the surface consisted of 100% sand, significantly greater forces were observed between three cushioned shock pad conditions. However, when rubber and sand was added to the surface at a ratio of 80% sand and 20% rubber, no significant differences could be observed between the same shock pads. As such, surfaces that have experienced compaction or have lost the rubber may be more greatly influenced by a more cushioned shock-pad. In the current investigation the amount of sand and rubber was recommended by the manufacturer to be 55% sand to 45% rubber (5:4). This is a greater proportion of rubber than the 80% sand, 20% rubber shown by McNitt et al., (2004) to prevent differences being observed between shock pads with mechanical tests.

In contrast to running, significantly larger peak impact forces were found on the less cushioned surface during turning. This ability to observe significant differences for peak impact force during turning is consistent with the previous chapter. This however, was only observable when impact force was measured with the pressure insole, and not with the force plate. This observation may show that pressure insoles are more sensitive to surface differences, possibly because it measures at the foot-shoe interface adding further evidence to support the opinion that pressure insole data provide greater sensitivity to boundary conditions (Dixon et al., 2008; Tillman et al., 2002; Dixon & Stiles, 2003; chapter 3). This finding however, is in contrast to those observed during running. It is possible that during the dynamic turning movements the forces are more influenced by smaller changes in surface cushioning, whereas if the aforementioned changes in the condition of the rubber crumb occur, differences in cushioning may also be evident during running.

As peak pressure at the first metatarsal was significantly reduced when turning on the more cushioned artificial surface, potential metatarsal damage may be reduced with this additional surface cushioning. It may be speculated that rather than a change occurring in response to the cushioning from the surface, this decreased loading may have been in response to a change in kinematics (Ford et al., 2006). However, observing that no differences in kinematic variables were present, the peak pressure differences may have been observed because of the more dynamic nature of the turn. This may cause the magnitude of the peak pressures at the first metatarsal to be greater than at other regions and so more influenced by the cushioning of the surface. The increased medial forefoot loading at metatarsal 1 on the harder surface is also consistent with the findings of Ford et al. (2006) observed during a cutting manoeuvre. This observation also supports previous evidence regarding the ability of pressure measurements to distinguish between surfaces of different cushioning.

The observation that both the impact force and first metatarsal peak pressure were greatest on the harder surface when turning, reaffirms the conclusions of the previous chapter regarding the importance of the contribution of mid- and fore-foot loading during the impact phase. The increased loading of these areas may contribute to the larger impact force observed during turning, assuming that this greater magnitude occurs at the same time as during running. This would explain why impact force increased significantly whilst heel forces did not, and why differences were shown during turning and not running. Having observed that there were no significant differences for the kinetic variables during the ‘braking step’, it can be concluded that the movement pattern observed between conditions was unchanged through this mechanism.

4.4.2. Footwear

It has been proposed that with the placement of additional material at the heel, the heel is raised relative to the forefoot (Clement et al., 1984; Leach et al., 1981) and the magnitude of peak dorsi-flexion is lowered during mid-stance (Dixon & Kerwin, 1999), reducing the loading on the Achilles tendon. This mechanism has been suggested to be behind the success of heel inserts to help treat and reduce the incidence of overuse injury to soft tissue such as the Achilles tendon (Fraunø et al., 1993; Nistor, 1981; MacLellan, 1984; MacLellan & Vyvyan 1981). However, the observation of no significant difference in peak dorsi-flexion when running and turning with the use of a 10 mm heel insert, suggests no such mechanism is apparent in the current investigation.

The observation in this study is similar to those made by Dixon and Kerwin (1999). In their study, 7.5 mm inserts did not significantly reduce peak dorsi-flexion during running. This however, conflicts with the injury treatment studies where much smaller heel inserts (<2.5mm) have been successful (Nistor, 1981). It is therefore suggested that the external measurements used in the present investigation are not sensitive enough to detect differences between footwear with and without heel inserts. The magnitude of internal force can be measured with in–vivo measurements (Komi, 1990), although this can be expensive, painful for the athlete and it can be difficult to attain ethical approval. Estimations of internal loads can be obtained using inverse dynamics or quasi-static techniques (Reinschmidt & Nigg, 1995; Scott & Winter, 1990). Since joint moments are sensitive to the magnitude of force and the length of the moment arm, small changes in each may have large effects on the magnitude of the joint moment (Winter, 1984). Likewise, Achilles tendon strain can be measured through ultrasound measurement and with biomechanical modelling (McGuigan et al., 2007; Dixon & Kerwin, 1999). These measurements may have been more appropriate and better able to distinguish significant differences between the heel lift conditions, detailing the mechanism behind the injury protection.

The success of heel inserts has also been suggested to result from reduced impact loading (Light & MacLellan, 1977; MacLellan & Vyvyan, 1981). Statistical analysis showed that in the current study there were no significant differences between the footwear conditions for any of the kinetic measurements except time to peak impact force which was significantly later with the heel insert when turning. This later time indicates a longer loading period which may cause a lower frequency of shock to travel the lower extremity (Whittle, 1999; Lees & McCullagh, 1984). As such, the structures do not experience a sudden loading. This may go some way to explaining the lower risk of injury with the visco-elastic heel lift experienced in past injury studies.

In previous research investigations, significant differences have been shown with Sorbothane insoles for peak rate of loading compared to a no insole (control) condition (Dixon et al., 2003; House et al., 2002; Windle, et al., 1999). The current study does not support these findings. It is also possible that GRF may not provide a good indicator of the impact shock waves that stress the lower extremity structures (Bobbert et al., 1992; Hamill, 1996). As the cushioning insole covers the entire plantar foot, it was suspected that this would significantly reduce peak pressures, peak pressure loading rates at the medial and lateral heel and the first and fifth metatarsals and the peak propulsive forces compared to the heel insert and control conditions. However, as with the previous measurements, the lack of significant differences in either peak propulsive force or peak pressures at the forefoot is also in contrast to past investigations (Dixon et al., 2003; House et al., 2002; Windle, et al., 1999). Because of the results in this investigation, even when surface- footwear interactions are taken into consideration, the claims of the Sorbothane manufacturer regarding the insole used in this investigation are still unsubstantiated. However, the cushioning provided by the surfaces may have been too high for differences to be shown between footwear types. Therefore, if a greater range of surface density was used, for example with a harder natural playing surface, significant differences may have been shown between footwear conditions.

It was highlighted in the introduction that by providing the soccer player with greater cushioning, the key design characteristic of a low profile soccer boot may be compromised, potentially increasing the risk of lateral ankle injury. Plantar flexion during turning was significantly greater with the heel insert. This would increase the loading of the lateral ankle ligaments (Fujii et al. 2005). However, the observation that rearfoot inversion was not significantly different suggests that the change in sagittal plane foot alignment did not influence the instability of the ankle joint. To potentially explain the lack of differences during turning, the kinetic measurement of the "braking foot" or the step made prior to the turn was used to assess whether the participants made a sudden change prior to the movement, which may have influenced the turning kinematics and kinetics. However, as no significant differences were observed during this step, it is speculated that the data collected during the turning movement are in response to the test conditions. The consistency in rearfoot angles between footwear conditions may therefore relate to increased protection from the peroneal muscles to prevent increased rearfoot movement. If during a turning movement the magnitude and rate of lateral forces are increased, or muscle pre-activation is reduced, the initiation of muscular support may not be sufficient or in adequate time to prevent extreme ankle movement from occurring (Hopkins et al., 2007). As such, injury may occur. This may happen if the surface traction is increased. Also, the inversion response during the turning movements may be accentuated if a more extreme movement velocity was used (Park et al., 2005) or if conditions were unanticipated, rather than predicted based on when they are told to turn. This may amplify the effect of any instability due to the heel insert so that participants could approach the thresholds of injury in game situations. The lack of significant difference in ankle inversion during turning may also relate to the fact that the use of external measurements fails to estimate the internal forces which are influenced by the magnitude and location of the ensuing forces and the movement of the joint. Small changes in these measurements can have a large effect on the measurement of joint moments (Winter, 1984). As such, to understand the loading of the ligaments, internal force estimations are needed

4.4.3. Conclusions

In summary, participants performing turning movements on the more cushioned shock pad experienced significantly reduced impact forces and peak pressures at the first metatarsal. This is agreement with the initial hypothesis that lower extremity loading would be reduced on the more cushioned shock pad, yet results collected during running did not support the hypothesis. The implication of these findings is that the shock pad density influences the cushioning provided to the soccer player, and therefore has potential to influence the risk of injury, particularly during turning. Consequently, it is suggested that surface manufacturers should use a more cushioned shock pad when designing third generation artificial surfaces. In addition, the finding that the response of the soccer player is dependent on the shock pad highlights that comparisons of artificial surfaces with natural turf should acknowledge that any conclusions are specific to the artificial surface type tested and cannot be assumed to be true for third generation surfaces in general. Also, the observation that significant differences are present with the measurement of peak impact force with the pressure insole, but not the force plate, supports previous suggestions that pressure insoles may be more sensitive to changes in surface cushioning than force plates, and thus should be utilised in future research investigations.

The comparison of the control and insole conditions with the heel insert did not show differences in dorsi-flexion during running and turning, which is in contrast to the hypothesis regarding this variable, and also the common belief that by using heel inserts this angle is reduced. The use of different footwear conditions showed no significant differences for peak propulsive force, peak pressure and peak pressure loading rate at the first and fifth metatarsal when running and turning, which is in contrast to the initial hypothesis of reduced fore-foot loading with the cushioning insole and increased loading with the heel insert. Likewise, peak impact force and peak rate of loading and peak pressure at the medial and lateral heel were not significantly different when running or turning which is in contrast to the hypothesis that for the control condition the values would be greater than for the heel insert and insole conditions. Observing that there was a significantly greater time to peak force with the heel insert when turning, suggests a potential method by which lower extremity injury may be reduced, yet indicating only partial support for the hypothesis that lower extremity loading is reduced. As most studies that detail successful intervention with the heel insert are during running tasks, and that no difference was observed for any biomechanical variable during this movement, the mechanism behind these injuries is still unclear.

The assessment of measurements associated with lateral ankle ligament damage during turning revealed that the participants experienced significantly greater plantar flexion with the heel insert. This would increase the loading on the lateral ankle, suggesting greater pain experienced from the lateral ankle structures. This finding supports the original hypothesis that plantar flexion would be significantly greater with the heel insert when turning, yet observing rearfoot inversion magnitudes that were not significantly different is in contrast to the original expectations. As such, the heel insert and the subsequent increase in plantar flexion did not seem to induce increased rearfoot inversion which characterises lateral ankle ligament damage. The increased plantar-flexion does, however, indicate potential for increased inversion if traction or movement speed were to increase. Further still, the findings of similar dorsi-flexion and inversion magnitudes, may relate to the fact that the use of external measurements are not good at indicating internal forces. As such, it is suggested that internal loading is considered in future research, particularly if the heel insert interventions are to be recommended for use in soccer.

5. Study Three

Understanding the effect of a heel insert intervention on the loading of the ankle joint and Achilles tendon during running and turning

5.1. Introduction

Achilles tendon injury is problematic in soccer, particularly during preseason, where nearly one third (32%) of Achilles related injuries are sustained (94% were either tendonitis or paratendonitis) (Woods et al., 2002). The susceptibility to Achilles tendon injury in soccer has been reduced with the use of commercially available heel inserts (Faunø et al., 1993; MacLellan, 1984). As such, heel inserts may prove a successful intervention to lower the risk of Achilles tendon injury during the preseason. The mechanism behind the reduction in injury with heel inserts is unclear, yet one theory relates to the visco-elastic properties of some heel inserts, (Light & MacLellan, 1977), whereas another is that heel inserts produce a change in the orientation of the foot, where the heel is raised relative to the forefoot (Clement et al., 1984; Leach, et al., 1981). This is suspected to reduce the maximum dorsi-flexion angle during the mid-stance phase of gait, which lessens the eccentric force applied to the tendon to control the movement. These factors were assessed with kinetic and kinematic data in chapter four of this thesis, with no significant differences in peak force or peak dorsi-flexion measurements being detected. One suggested reason for these results is that GRF does not faithfully reflect the specific force at the foot (Hamill, 1996). In addition, the measurement of kinematic variation does not provide the internal muscular forces which are sensitive to small changes in GRF and moment arm length (Winter, 1984).

To quantify the mechanics behind the success of heel lift interventions in the prevention and treatment of Achilles tendon injury, authors have calculated the peak plantar flexion muscle moment. This muscle moment represents the net force occurring during dorsi-flexion to control the movement during mid-stance. It is assumed that during dorsi-flexion, between 85-100 % of the net force is contributed by the triceps surae muscle group and is applied via the Achilles tendon (Scott & Winter 1990; Winter, 1980). Thus, this measurement has been used as an indicator of Achilles tendon force (Reinschmidt & Nigg 1995), although this does not directly estimate this force. Dixon and Kerwin (2002) presented a modelling method which aimed to provide a reliable, subject specific estimate of the forces actually occurring in the Achilles tendon. Despite these non-invasive indicators of Achilles tendon force being used, neither Reinschmidt and Nigg (1995) (2.1-3.3 cm thickness heel lift) nor Dixon and Kerwin (2002) (7.5-15 mm thickness heel lift) could produce distinguishable Achilles tendon forces when the heel was raised by different amounts. It has been suggested that the use of peak force is not always sufficiently sensitive to distinguish statistical differences between risk factors (Dixon & Kerwin, 2002). Instead, the rate at which this internal force is being applied may be more able to show differences between heel insert conditions. Dixon & Kerwin (2002) calculated the peak and average rate of loading of the Achilles tendon force whilst running in a 7.5 mm heel insert, a 15 mm heel insert, and no heel insert condition. Using the peak loading rate, no significant differences were shown between the two heel insert conditions or the control, whereas the 15 mm heel insert significantly reduced the average loading of Achilles tendon force. Consequently, the measurement of average loading rate may be more suitable for determining differences between heel insert conditions and potentially explain the reduced risk of injury associated with wearing a heel insert. One limitation of the methodology developed by Dixon and Kerwin (2002) was the use of two-dimensional sagittal plane data when estimating Achilles tendon force. Further improvement of the Achilles tendon force calculation method may be obtained by using three dimensional moments and moment arm data. Further still, the use of soccer specific footwear may prevent atypical running gait experienced during barefoot trials performed in previous investigations (Johansson et al., 2006). In turn, this could enable improved measurement sensitivity and subsequently aid with the observation of significant differences between conditions for both absolute magnitude and average loading rate of Achilles tendon force with heel inserts.

A further reason for the lack of differences between with and without heel lift in previous studies may relate to the great variability that exists between athletes in response to a change in surface or footwear conditions (Bates et al., 1983). Reinschmidt and Nigg (1995) reported that single participant analysis did show for two of their five participants, a significant reduction in plantar flexion moment with the use of a heel insert. This suggests that, assuming no lengthening in moment arm length between joint centre of rotation and Achilles tendon line of action, a decrease in joint moment may for some at least, indicate a mechanism behind the reduction of injuries. Dixon and Kerwin (2002) found that lower Achilles tendon forces were experienced in some heel toe runners although these differences were not statistically analysed. Further still, Dixon and Kerwin (1998) found that peak Achilles tendon force of a participant characterised as having a heel-toe running pattern, increased with the use of a heel insert. As such, this may further advocate the necessity for single subject analysis, as in some situations the behaviour change in a minority of participants may obscure the change in the majority when group analysis is performed. As a consequence, the mechanism behind the lowering of injury in most participants may be disguised.

Although heel inserts seem to provide some benefits in the reduction of injury, the previous chapter discussed the potential problems that may also exist with their use. It was highlighted that the use of the heel insert may decrease the stability of the ankle joint during turning movements and possibly contributes further to the risk of an ankle injury particularly to the ATFL. In the previous chapter, the heel insert inclusion in the soccer boot significantly increased the plantar flexed position of the foot, which could cause greater force to occur on the ATFL. Reinschmidt and Nigg (1995) found that during running the magnitude and time of occurrence of the peak dorsi flexion moments (Figure, 5.1) were significantly increased with a greater heel height. Although this was performed during running, a greater dorsi-flexion moment and average loading rate may be evident during turning. As such, this change in moment would indicate greater internal force to control the plantar flexion and thus would be consistent with the change in peak plantar flexion observed in the previous chapter. This may increase the strain experienced by the ATFL, increasing the risk of injury. Also, to limit the risk of inversion injury, soccer boots are designed to be low to the ground with very little cushioning at the heel. However, the additional thickness provided by the visco-elastic material of the heel insert may not only cause the rear-foot to become more unstable and experience increased inversion angulations as a result of the raised ankle plantar flexion, but also increase due to the lateral edge of the insert being more easily compressed than the boot alone. Despite this suggestion, there has been no research data to indicate the effect of heel lift interventions on lateral stability during turning. As such, the peak eversion moment and average eversion moment loading rate, may contribute to representing the loading characteristics occurring within the lateral ankle joint when performing inversion movements.

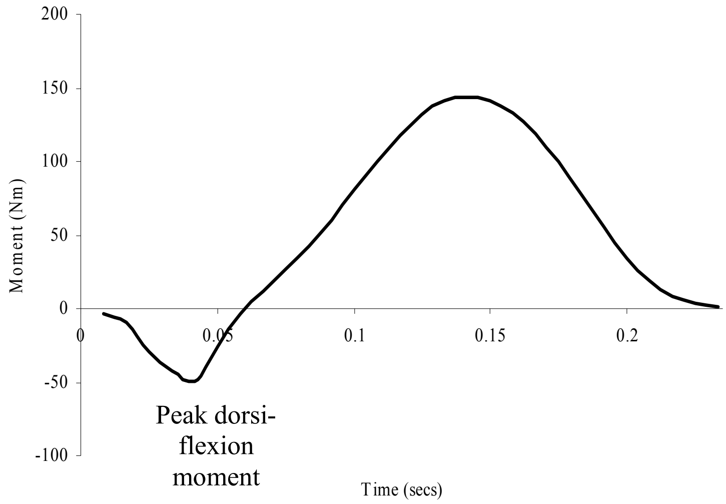


Figure 5.1. A typical moment-time history of dorsi flexion and plantar flexion during running.

5.1.1 Aims and objectives

The aim of the present investigation is to assess the influence of a commercially available heel insert on the peak plantar flexion moment and Achilles tendon force during running and turning and the average loading rates of these measurements. In addition, the investigation will assess the risk of inversion injury during turning in response to the heel insert by measuring the peak inversion and dorsi-flexion moment and average loading rates of these moments.

It is hypothesised that the magnitude of peak plantar flexion moment and peak Achilles tendon force, and the average rate of loading of these measurements, will be significantly reduced with the inclusion of the heel insert in a soccer boot when running and turning. It is also hypothesised that the inclusion of a heel insert will significantly increase rearfoot eversion and dorsi-flexion moments and average loading rates of each of these variables. This would indicate a greater risk of ankle inversion injury whilst wearing the heel insert, thus questioning their appropriateness for use in soccer.

5.2. Method

5.2.1. Participants

Nine male soccer players (83.4 kg [S.D. 5.8], 23 yrs [S.D. 3.7], Achilles tendon radius 19.13 cm [S.D. 2.3], ankle width 0.072 cm [S.D. 0.005], forefoot width 0.10 cm [S.D. 0.005], size 10 feet) participated in the research investigation. All participants regularly participated in soccer and had recent experience of playing on a third generation artificial surface. Each participant also had a heel-toe running gait and was injury-free throughout and three months prior to the time of the data collection. Participants were asked to mention any feelings of discomfort that came from the tasks involved. All participants were made aware of the aims and objectives of the research investigation and were told that they were free to withdraw from participation for any reason and at any time. To confirm that each participant was aware of the nature of the investigation, information sheets were provided and signed consent forms were gathered. The project was approved by the ethics committee within the School of Sport and Health Sciences, University of Exeter.

5.2.2. Data collection

To calculate the moments occurring during running and turning, both kinematic and kinetic data were collected. For the collection of kinematic data, each participant was required to have reflective markers placed on the hip, knee, lateral maleolus, fifth metatarsal, shin and toe (Figure 5.2). Two markers were also placed on the heel to define the line of the calcaneous (Cal 1 and Cal 2) and two more to define the line of the Achilles tendon (Ach 1 and Ach 2) (Figure 5.3). The most proximal of these was positioned at the same height as the shin marker (Figure 5.2). The position of these markers was based on research conducted by Soutas-Little et al. (1988) and was used to establish local reference planes for the calculation of three-dimensional kinematics.

Figure 5.2. An example of marker placement from the sagittal view

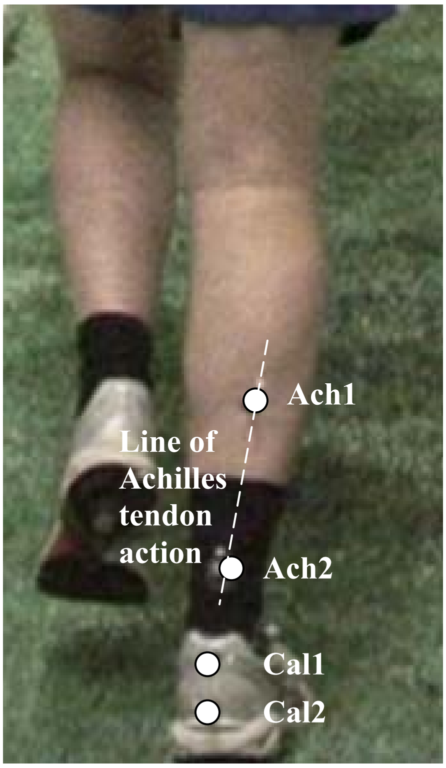


Figure 5.3. An example of marker placement from the posterior view

An eight camera (Pulnix, TM-6703 progressive scan, 120 Hz) (Figure 4.10) automatic tracking set up was used (Vicon, Motus version 6.1, Englewood, CO, USA). Each camera was distributed around the calibration volume and positioned along the length of a biomechanics laboratory, with each focused on the force plate (Figure 5.4). The cameras were genlocked to enable synchronised three-dimensional coordinates to be obtained for each marker worn by the participants. Marker co-ordinates were calculated via the application of a direction linear transformation (DLT).

Environmental set up

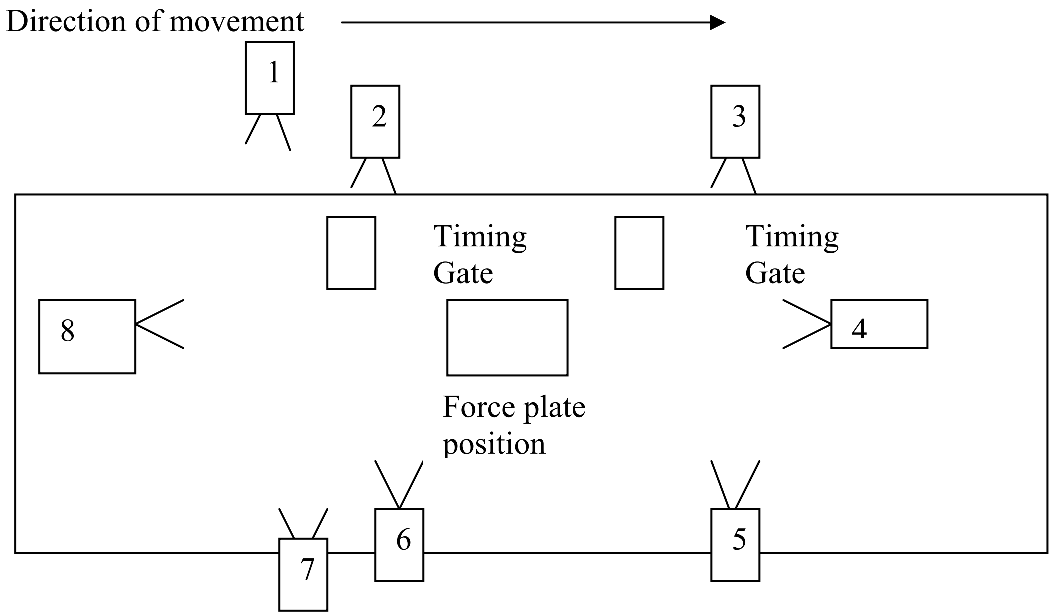


Figure 5.4. Schematic of the laboratory set up including the timing gates, cameras and force plate. Cameras are represented by the boxes marked 1-8

To calculate the three-dimensional moments about the ankle joint, ankle joint and forefoot centres were required. These were calculated by measuring the ankle and forefoot width for each participant. To determine ankle width, the procedure used a calliper to determine the distance from the lateral maleolus to the medial maleolus (Figure 5.5). The calliper was also used to measure the distance between the first and fifth metatarsal at the widest location to measure the forefoot width (Figure 5.6).

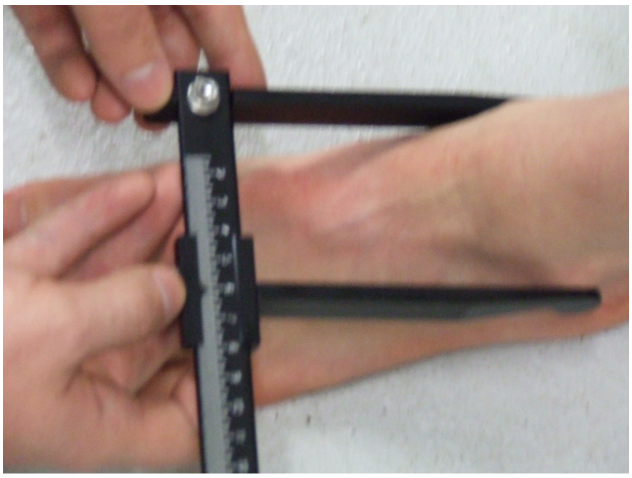


Figure 5.5. An example measurement of the ankle width using a calliper

Figure 5.6. An example measurement of the forefoot width using the calliper

Using these values, the co-ordinates of the ankle and forefoot centres were calculated for each participant by using Equation 1 and Equation 2. Shoe material thickness was regarded as negligible when determining forefoot width.

Ankle joint centre = L + ((0.5 * A) + R)* X Equation 5. 1

Forefoot centre = F + ((0.5* W) + R) * X Equation 5. 2

Where

L= location of the lateral maleolus marker

A= ankle width

R= Radius of marker

X= X foot (see appendix E)

F= location of the Fifth metatarsal marker

W = forefoot width

Quintic splines (Woltring, 1985) were fitted to the raw coordinates to obtain smooth continuous time histories and first and second derivatives. Acceleration data were calculated by the first central difference method. The coordinates of each marker were used to define the ankle and foot orientation as well as the ankle and rearfoot angle configurations during the ground contact phase.

To collect kinetic data, specifically force (Fx, Fy, and Fz), centre of pressure (ax, ay) and free moment (Fm), a force plate (AMTI, 960 Hz) was used. This was positioned approximately mid-way along the length of a biomechanics laboratory.

The calculation of the three-dimensional moment used an inverse dynamics technique. This required knowledge of foot moment of inertia, mass and COM. The foot mass and COM were calculated using adult male cadaver data from Clauser et al. (1964) and moment of inertia data using the data provided by Whitsett (1963).

The synchronous smoothed coordinate and force data were transferred from the Vicon Motus software into a Matlab program (Matlab, 7.0.4, TheMathsWorks, USA). Within the Matlab program, a code was written that interpolated the 960 Hz kinetic data to 120 Hz. The kinetic data were interpolated rather than the kinematic data extrapolated from 120 Hz to 960 Hz. This was because Stiles (2005) found that the noise from extrapolated kinematic data was magnified to unacceptable levels at the second and third derivatives. Therefore, this data is not appropriate for use in an inverse dynamics equation. The calculation of three-dimensional moments occurring during plantar flexion, dorsi-flexion and inversion movements were performed using three dimensional inverse dynamics equations written within a Matlab code based on previously published methods (Kawamoto, Ishige, Watarai & Fukashiro, 2002; Kwon3d, 2009) (Appendix, E).

During the movements, the conventions of the calculated muscle moments were that a negative moment represented a resistance to extension of the segment. On the other hand, a positive movement represented a resistance to flexion (Figure 5.7).

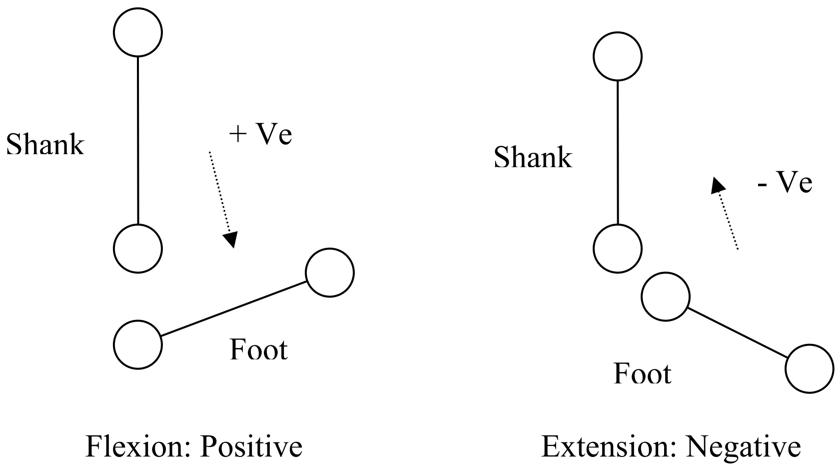


Figure 5.7. A schematic of the moment conventions that describes the direction of the muscle moment of foot-shank segments.

5.2.3. Achilles tendon loading

Three-dimensional Achilles tendon force was calculated by adopting a similar technique to that developed by Dixon and Kerwin (1998; 2002). In their study, the perpendicular two-dimensional moment arm distance from the ankle joint centre of rotation to the line of the Achilles tendon was used. In the present study a three-dimensional moment arm was required and this was calculated as the distance between the ankle joint centre and the line of the Achilles tendon represented by the two markers on the back of the shank (Figure 5.3).

Dixon and Kerwin (2002) indicated that the length on the moment arm was influenced by the skin thickness that surrounds the tendon sheath, and the radius of the external marker size used to define the line of the Achilles tendon. To account for these additional distances, the radius of the marker, as well as the skin covering the Achilles tendon, was removed from the moment arm length prior to the calculation of Achilles tendon force. To calculate the skin thickness, the radius of the Achilles tendon width at approximately 5 mm from the Achilles tendon insertion point was measured with a calliper for each subject. This was scaled using the skin thickness-Achilles tendon ratio which Dixon (1996) reported as 3.9 mm when the Achilles tendon radius was 7 mm.

5.2.4. Test conditions and task

Fifteen metres of third generation artificial shock pad (Arpro® Expanded polypropylene, 24mm ± 0.5mm thick, Brock International) which had a measured manufactured density of 65g and a mechanical hardness of 1254.3 N (S.D. 48.5 N), was laid across a concrete laboratory floor (see Table 4.1). Placed upon the shock pad was a third generation turf of similar length (Astroplay MXS 40, Lano sports, Herelbeke, Belgium). Upon this, 10 kg/m^2 of sand and 8kg/m^2 of rubber crumb (5:4 ratio of sand to rubber) were distributed as recommended by the manufacturer. The force plate was positioned underneath, and approximately in the centre of the surface in both width and length. This was represented by a square marked on the surface.

Participants performed 10 running and 10 turning movements upon the artificial turf surface. These movements were performed whilst wearing a soccer boot with a moulded stud configuration (Adidas, Copa Mundial) (Figure 5.8) with (experimental) and without (control) a commercially available 10 mm Sorbothane heel insert (Sorbothane

Shock Stopper, Sorbopro, Layland, Lancashire, UK) (Figure 5.9). These conditions were tested in random order for each participant.

Figure 5.8. An example of Adidas Copa Mundial soccer boots

Figure 5.9. An example of 10 mm Sorbothane heel insert

The participants were asked to run the length of the artificial turf surface at a speed of 3.81 $m.s^{-1}$ (± 5%), monitored by photosensitive timing gates positioned one meter either side of the force plate. Participants were asked to run at this speed and place their right foot within the box marked on the surface without changing their normal stride pattern and then continue running to the end of the artificial turf surface. Participants were also asked to perform turning trials where they were to run at a comfortable speed up to the force plate and place their right foot into the marked box. They were then asked to twist their hips whilst the foot was fixed to the surface and push off in the direction they came. The speed of the turn was monitored by timing gates, although no set speed was used. Instead, the speed of the first comfortable turn was used for the remaining turning trials. This made the selected speed specific for each participant. All trials that were not performed as directed or were not at the correct speed were subsequently repeated.

5.2.5. Statistical analysis.

Peak plantar flexion moment and Achilles tendon force were determined from the calculated moment data during running. Likewise, peak plantar flexion moment and Achilles tendon force were identified for turning, along with the peak eversion and dorsi-flexion moments. Average loading rate for each of these values was calculated by dividing the peak moment/Achilles tendon force by the time over which it occurred. The mean of the 10 trials collected for each condition were statistically compared using a paired samples t-test. Normality of data distribution was tested by determining skewness and kurtosis statistics. This ensured that the data met the assumption of normal distribution, necessary for a paired samples t-test to be performed. Individual participant data was analysed by performing paired samples t-tests between each of the 10 trials from either condition for each participant independently. The alpha level was set at 0.05 for all tests. Statistical tests were performed using SPSS (15.0 for Windows).

5.3. Results

5.3.1. Running

Table 5.1 demonstrates the results of the experimental procedures when running. Peak plantar flexion moment was compared between the soccer boot condition with and without a 10 mm heel insert. No significant difference was shown for this variable nor where there any significant differences observed for the measurement of the average plantar flexion moment loading rate between the heel insert conditions. No significant differences were also shown between conditions for the measurements of peak Achilles tendon force and average Achilles tendon loading rate.

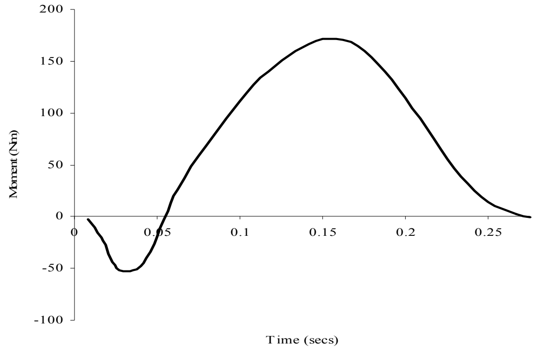
Table 5.1 *Results of comparison whilst running in soccer boots with and without a heel insert*

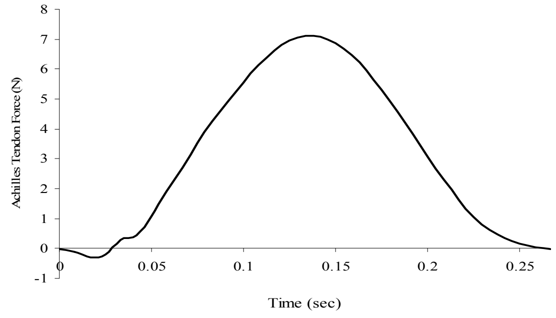
		Mean	P
Peak plantar flexion moment (N)	Soccer boot	196.9 (S.D. 73.5)	0.6
	Soccer boot + Heel insert	192.5 (S.D. 53.3)	8
Average plantar flexion moment loading rate ($Nm.s^{-1}$)	Soccer boot	1612.2 (S.D. 526.6)	0.9
	Soccer boot + Heel insert	1617.4 (S.D. 795.7)	7
Achilles tendon force (N)	Soccer boot	5357.5 (S.D. 2378.2)	0.26
	Soccer boot + Heel insert	5907.4 (S.D. 3199.2)	
Achilles tendon force (BW)	Soccer boot	6.6 (S.D. 3.0)	0.25
	Soccer boot + Heel insert	7.3 (S.D. 4.0)	
Average Achilles tendon force loading rate ($N.s^{-1}$)	Soccer boot	255715.5 (S.D. 109306.8)	0.92
	Soccer boot + Heel insert	259396.3 (S.D. 152438.3)	

Figure 5.10 shows sample time histories for dorsi flexion and plantar flexion moment, and Achilles tendon force for each subject. During running, the time-histories for each participant tended to show similar trends for the experimental and control conditions. Participants typically experienced an increase in dorsi-flexion moment during initial foot contact, followed by an increase in plantar flexion moment during mid-stance. However, some participants did not always experience a dorsi-flexion moment. Similar characteristics were observed for the time history of the Achilles tendon force.

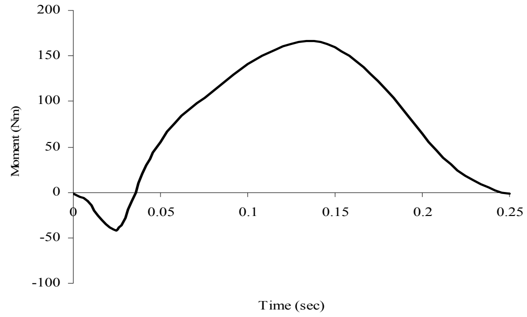
Participant 1

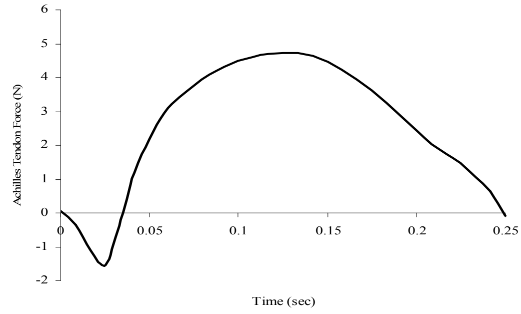
Heel insert





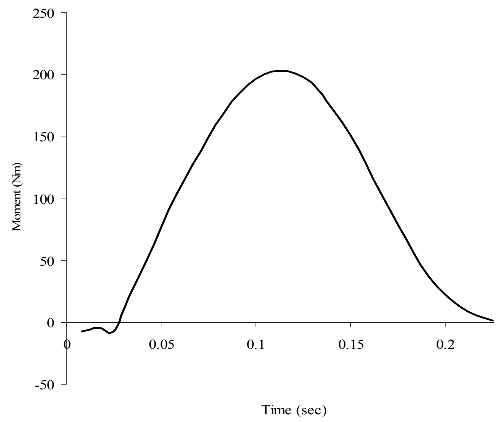
Without heel insert

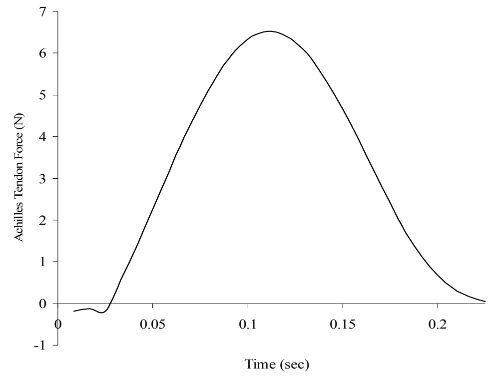




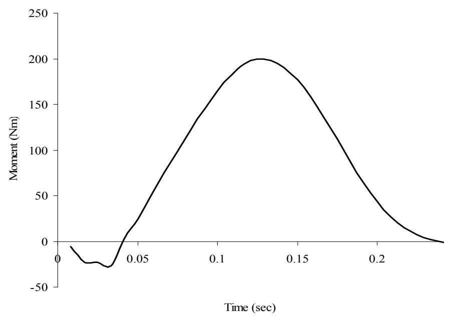
Participant 2

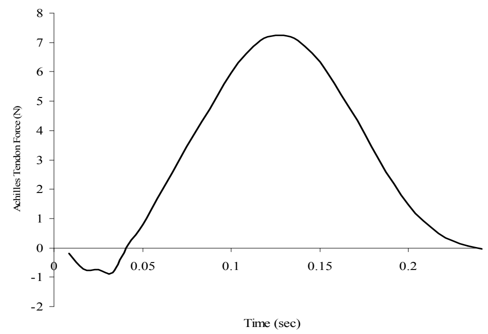
With heel insert





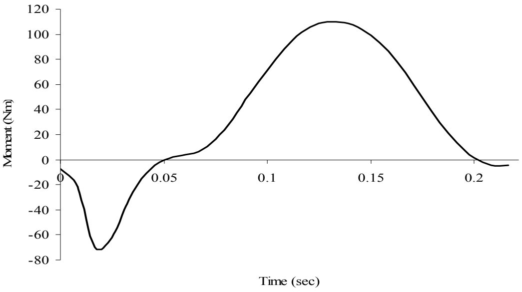
Without heel insert

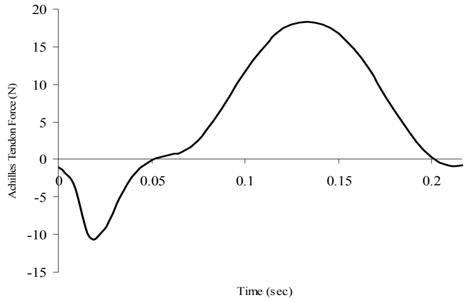




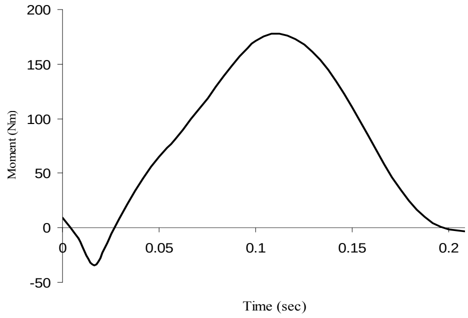
Participant 3

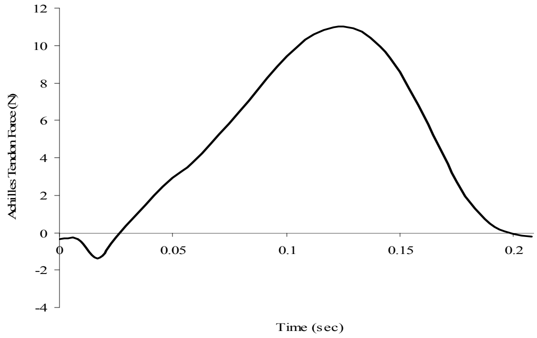
With heel insert





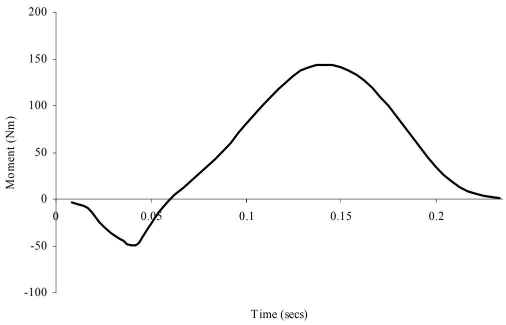
Without heel insert

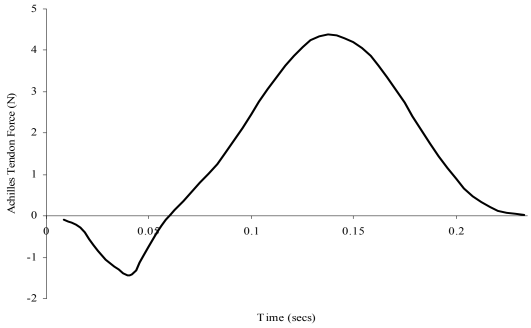




Participant 4

With heel insert





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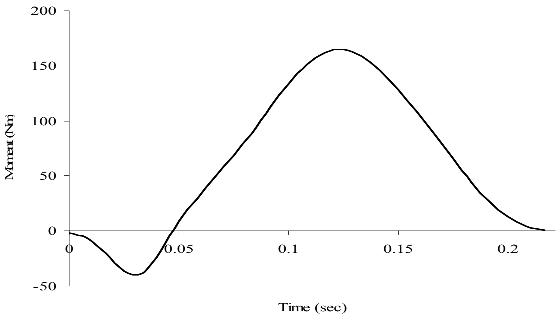
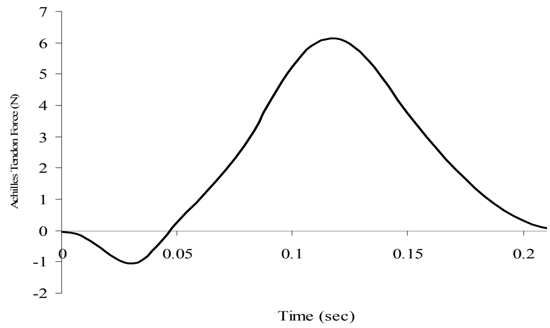
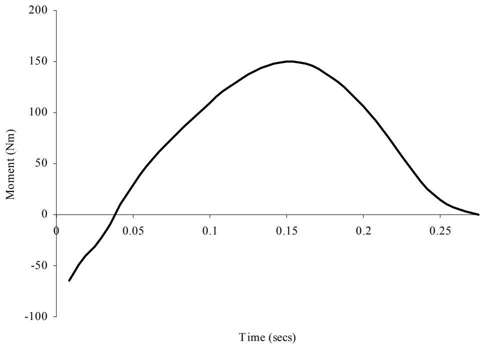
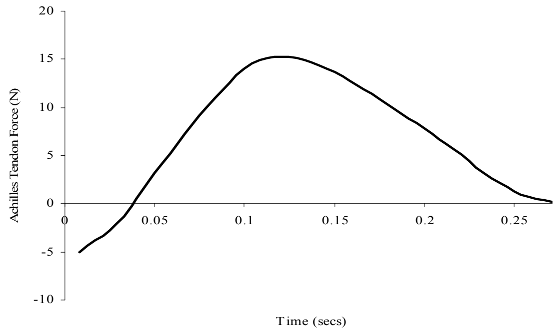




Participant 5

With heel insert





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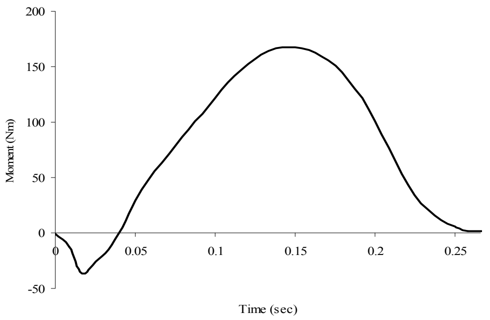
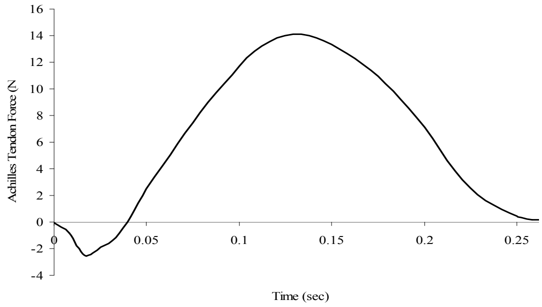
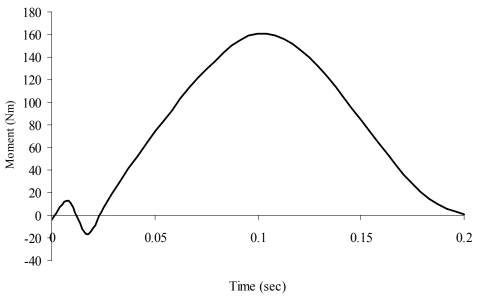
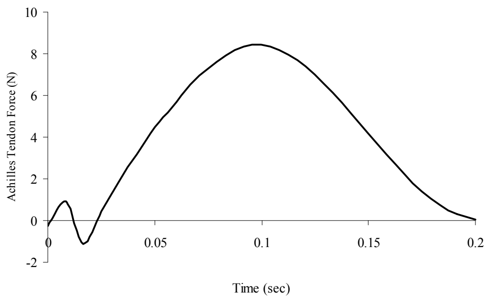




Participant 6

With heel insert





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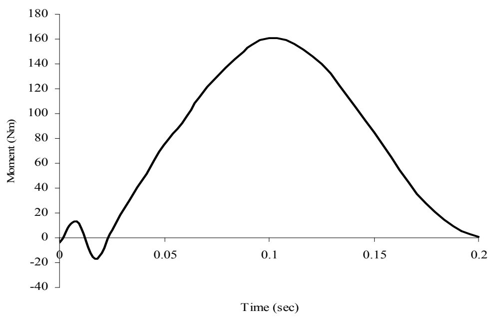
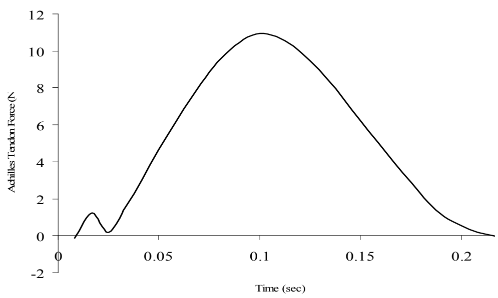
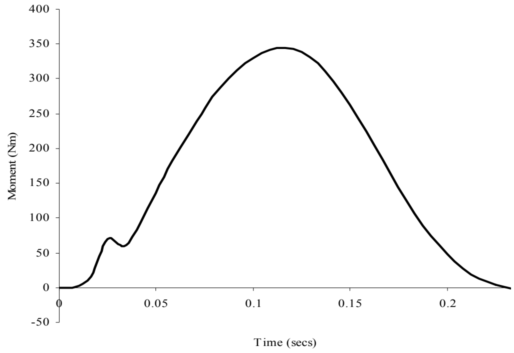
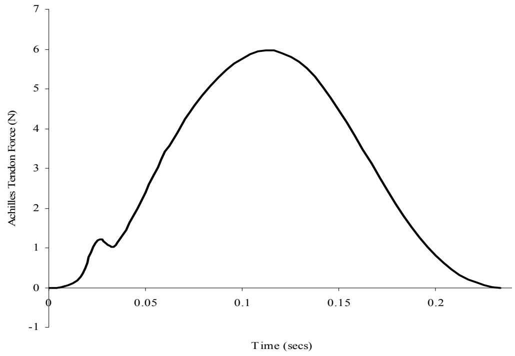




Participant 7

With heel insert





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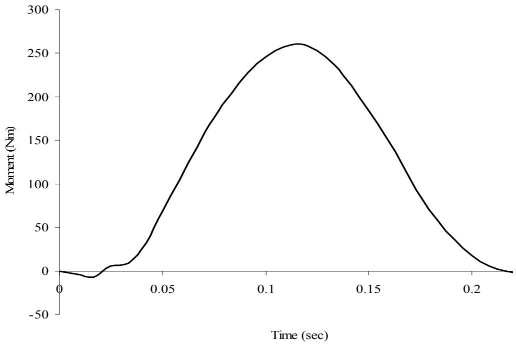
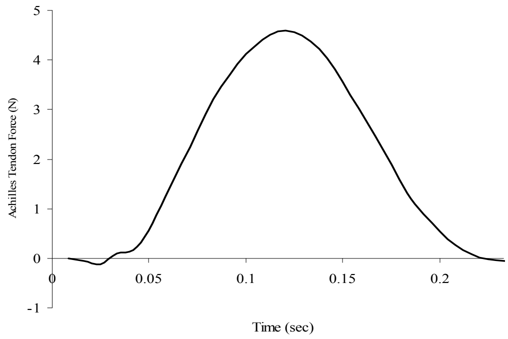
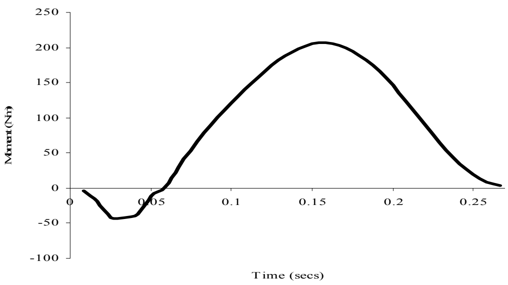
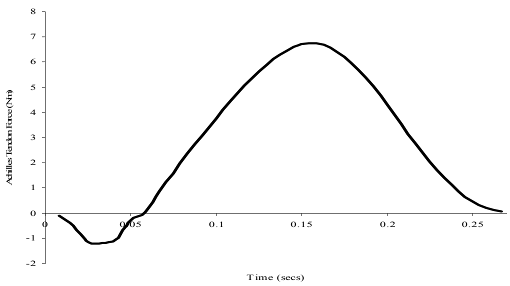




Participant 8

With heel insert





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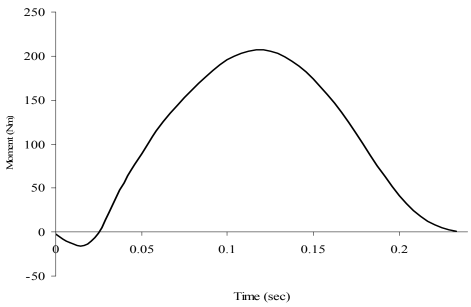
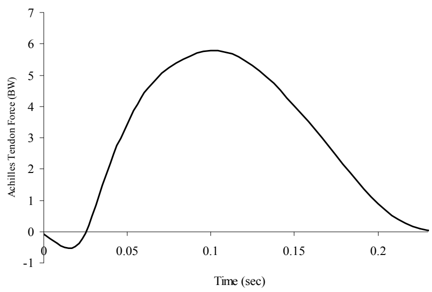
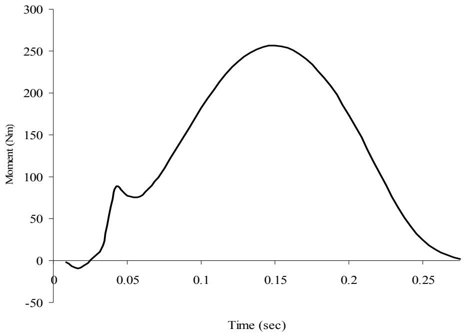
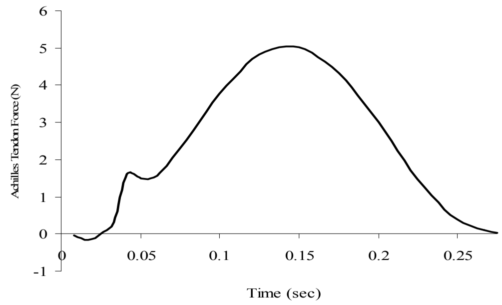




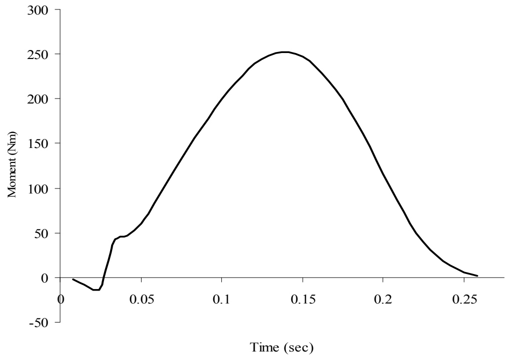
Participant 9

With heel insert





Without heel insert



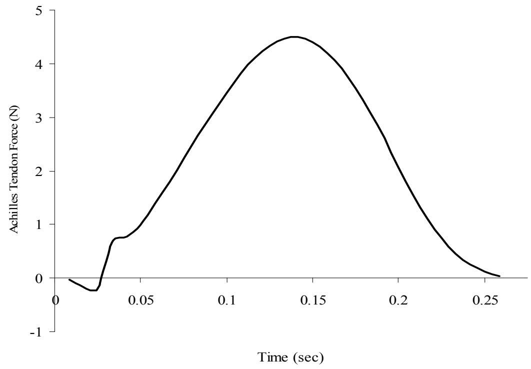


Figure 5.10. Profile of dorsi flexion and plantar flexion moment, and Achilles tendon force time histories for each participant

Table 5.2 shows the means of the individual data collected whilst running. Peak plantar flexion moment was significantly different for two participants, one of which was significantly greater (participant 7) and the other was significantly smaller (participant 6) with the heel insert. The average loading rate was different for four participants, three of which showed significant reductions with the heel insert (participants 3, 5, 6), whereas the other was significantly greater with the insert (participant 7). The differences in the peak Achilles tendon force between the footwear conditions were significantly different for three participants (participants, 1, 7, 8), with one participant showing a significant increase in peak Achilles tendon force (participant 7). No participants exhibited a significant reduction or increase in average Achilles tendon loading rate during the running movement.

Table 5.2 *Individual participant data whilst running with and without a heel insert in a soccer boot*

Participant		Peak plantar flexion moment (Nm)	Average rate of loading of plantar flexion moment ($Nm.s^{-1}$)	Peak Achilles tendon force (N)	Peak Achilles tendon force (BW)	Average rate of loading of Achilles tendon force ($N.s^{-1}$)
1	Soccer boot	185.4 (S.D. 7.58)	1488.2 (S.D. 115.2)	5137.1 (S.D. 442.1)	6.0 (S.D. 0.5)	395191.8 (S.D. 383361.7)
1	Soccer boot + Heel insert	180.7 (S.D. 9.13)	1471.5 (S.D. 190.2)	4434.6 (S.D. 920.9)	5.2 (S.D. 1.1)	185020.3 (S.D. 42570.5)
		0.24	0.82	**0.04***↓	**0.04***↓	0.12
2	Soccer boot	196.3 (S.D. 32.6)	1716.4 (S.D. 392.6)	5581.4 (S.D. 1282.6)	7.7 (S.D. 1.8)	258925.6 (S.D. 78002.3)
2	Soccer boot + Heel insert	197.2 (S.D. 23.7)	1677.6 (S.D. 315.4)	5397.7 (S.D. 934.6)	7.5 (S.D. 1.3)	249610.5 (S.D. 61487.1)
		0.94	0.82	0.73	0.79	0.72
3	Soccer boot	99.9 (S.D. 12.9)	792.92 (S.D. 98.33)	3484.9 (S.D. 764.7)	4.2 (S.D. 0.9)	279353.7 (S.D. 53251.9)
3	Soccer boot + Heel insert	82.0(S.D. 11.6)	619.6 (S.D. 89.0)	4680.0 (S.D. 2586.9)	5.6 (S.D. 3.1)	292870.3 (S.D. 195227.1)
		0.07	**0.01***↓	0.19	0.84	0.19
4	Soccer boot	145.1 (S.D. 28.0)	1069.7 (S.D. 228.9)	2715.8 (S.D. 936.8)	3.7 (S.D. 1.3)	115178.2 (S.D. 64703.0)
4	Soccer boot + Heel insert	149.8 (S.D. 36.3)	1061.7 (S.D. 303.7)	2834.4 (S.D. 830.0)	3.9 (S.D. 1.1)	133072.6 (S.D. 35402.8)
		0.78	0.95	0.79	0.51	0.79

5	Soccer boot	166.0 (S.D. 5.2)	1215.1 (S.D. 118.3)	7802.6 (S.D. 3125.39)	10.3 (S.D. 4.1)	331587.6 (S.D. 132840.4)
5	Soccer boot + Heel insert	160.9 (S.D. 11.8)	1036.2 (S.D. 174.4)	8440.9 (S.D. 1516.3)	11.1 (S.D. 2.0)	306980.4 (S.D. 41459.0)
		0.23	**0.02*↓**	0.57	0.59	0.57
6	Soccer boot	184.9 (S.D. 10.8)	2047.8 (S.D. 121.9)	9780.9 (S.D. 5584.9)	11.9 (S.D. 6.8)	402850.4 (S.D. 280490.1)
6	Soccer boot + Heel insert	181.9 (S.D. 9.2)	1754.4 (S.D. 132.5)	13388.8 (S.D. 7471.3)	16.2 (S.D. 9.1)	632358 (S.D. 362511.0)
		0.05*↓	**0.01*↓**	0.29	0.18	0.29
7	Soccer boot	261.5 (S.D. 15.3)	2472.2 (S.D. 253.2)	2584.3 (S.D. 170.9)	3.2 (S.D. 0.2)	127318 (S.D. 12540.89)
7	Soccer boot + Heel insert	347.8 (S.D. 48.5)	3446.8 (S.D. 447.9)	3764.8 (S.D. 1559.4)	4.7 (S.D. 1.9)	152593.6 (S.D. 67164.7)
		0.01*↑	**0.001*↑**	**0.04*↑**	**0.04*↑**	0.27
8	Soccer boot	231.8 (S.D. 7.8)	1746.2 (S.D. 141.29)	6384.2 (S.D. 1577.5)	7.5 (S.D. 1.9)	243847.4 (S.D. 58046.7)
8	Soccer boot + Heel insert	227.3 (S.D. 13.7)	1763.3 (S.D. 326.3)	5469.0 (S.D. 741.9)	6.4 (S.D. 0.9)	214208.1 (S.D. 46368.4)
		0.41	0.89	**0.14*↓**	**0.14*↓**	0.25
9	Soccer boot	261.3 (S.D. 29.9)	1961.4 (S.D. 302.4)	4746.3 (S.D. 568.3)	5.1 (S.D. 0.6)	147187.2 (S.D. 22672.3)
9	Soccer boot + Heel insert	244.9 (S.D. 18.1)	1725.6 (S.D. 245.0)	4756.5 (S.D. 538.6)	5.1 (S.D. 0.6)	167852.6 (S.D. 28968.6)
		0.25	0.13	0.78	0.78	0.61

↑ = increase with heel insert, ↓ with heel insert

Data are reported as mean and standard deviation. Where, * denotes a significant difference at the $p<0.05$ level

5.3.2. Turning

Table 5.3 presents the mean peak plantar flexion, eversion and dorsi-flexion moments and Achilles tendon force along with the average loading of each measurement, which were calculated with data obtained whilst turning. No significant differences were observed between the heel insert conditions for peak magnitude or average loading rate of the plantar flexion, dorsi-flexion, and rearfoot eversion moments or Achilles tendon force.

Table 5.3 *Table showing peak plantar-flexion moment, peak Achilles tendon force, peak eversion moment and average loading rate measurements whilst turning in soccer boots with and without a heel insert*

		Mean	P
Peak Plantar flexion moment (Nm.)	Soccer boot	116.0 (S.D. 24.1)	0.13
	Soccer boot + Heel insert	96.1 (S.D. 34.8)	
Average plantar flexion moment loading rate ($Nm.s^{-1}$)	Soccer boot	807.2 (S.D. 522.0)	0.09
	Soccer boot + Heel insert	480.2 (S.D. 245.6)	
Peak dorsiflexion moment (Nm)	Soccer boot	40.0 (S.D. 33.5)	0.67
	Soccer boot + Heel insert	36.0 (S.D. 26.9)	
Average dorsiflexion moment loading rate ($Nm.s^{-1}$)	Soccer boot	499.7 (S.D. 616.3)	0.45
	Soccer boot + Heel insert	865.5 (S.D. 1285.9)	
Peak eversion moment (Nm)	Soccer boot	-45.3 (S.D. 27.2)	0.95
	Soccer boot + Heel insert	-46.4 (S.D. 36.4)	

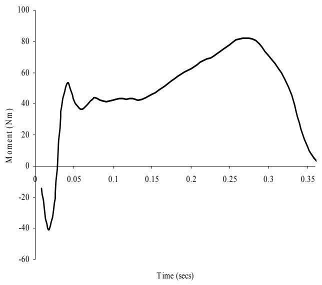
Average eversion moment loading rate ($Nm.s^{-1}$)	Soccer boot Soccer boot + Heel insert	-3491.0 (S.D. 3224.4) -3421.8 (2176.2)	
Achilles tendon force (N)	Soccer boot Soccer boot + Heel insert	4090.4 (S.D. 1826.4) 2880.2 (S.D. 1272.7)	0.07
Achilles tendon force (BW)	Soccer boot Soccer boot + Heel insert	5.0 (S.D. 2.2) 3.6 (S.D. 1.6)	0.07
Average Achilles tendon force loading rate ($N.s^{-1}$)	Soccer boot Soccer boot + Heel insert	119754.3 (S.D. 69202.5) 74862.6 (S.D. 69540.8)	0.13

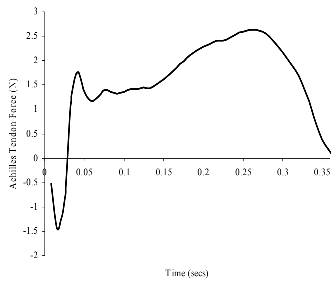
Figure 5.19 presents sample individual time history for dorsi-flexion/plantar flexion and eversion moment and Achilles tendon force experienced during turning. There were some common similarities in the time-histories between participants and conditions. The dorsi and plantar flexion moments showed that most participants exhibit a dorsi flexion moment that peaked early in the stance phase and then transferred into a plantar flexion moment during mid-stance. Some participants landed with a peak dorsi-flexion moment and increased into a plantar flexion moment immediately. The plantar flexion moment commonly peaked twice for most participants, although some variation in the overall shape of the curve was shown. The common time-history for the Achilles tendon force was similar to the ankle moment but variations in the shape were also shown for this measurement between participants.

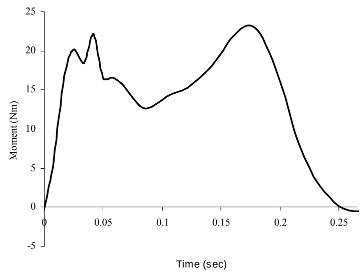
The measured inversion-eversion moment curve was commonly double peaked. The first peak represented the forces that occur in response to the need to control rearfoot inversion during impact. The foot subsequently performs an eversion movement during mid-stance which causes a reduction in the moment measured. Most participants continued to experience an eversion moment, although participants 2 and 9 exhibited an inversion moment. As the participant moves in to the propulsive period there is a re-inversion that causes an increase in the eversion moment measured. The peak value was typically the first, but occasionally this was not the case. However, the magnitude of the first peak was taken for comparison as this indicated the instability of the rearfoot due to the insert and the second peak (or third) represents the re-inversion of the rearfoot when the heel is not planted and therefore the size of the moment does not reflect any change in the instability. For one participant however, there was no inversion moment and therefore this participant (participant 9) could not be used to compare the inversion moment between conditions. For all measurements, the magnitudes of the various aspects were specific to the individual.

Participant 1

With heel insert







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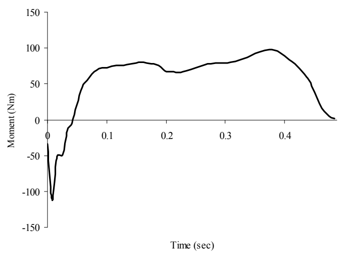
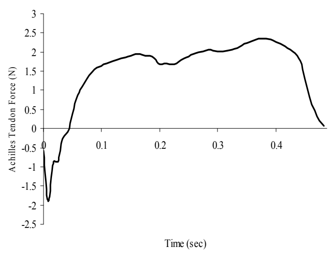
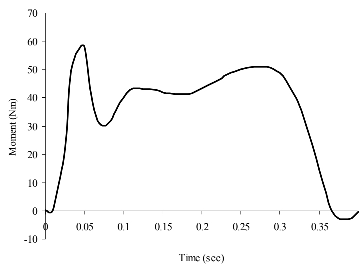
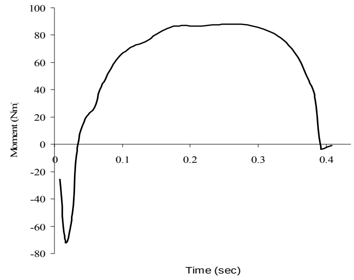


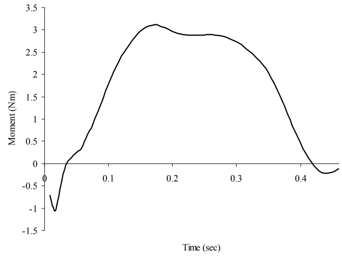


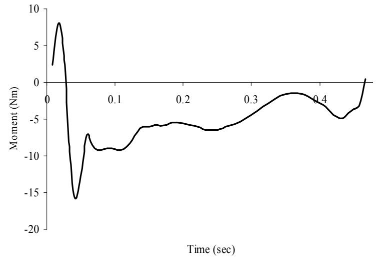


Participant 2

With heel insert







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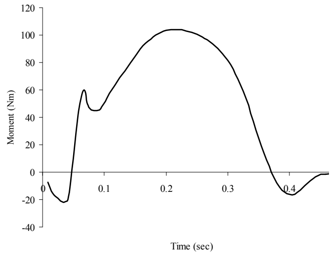
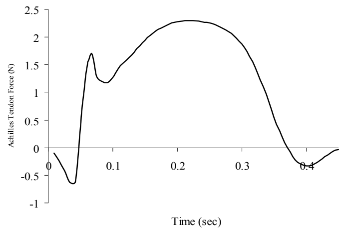
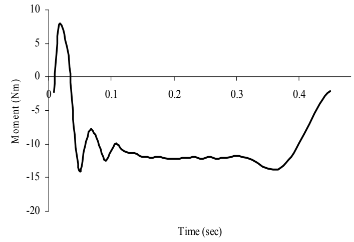
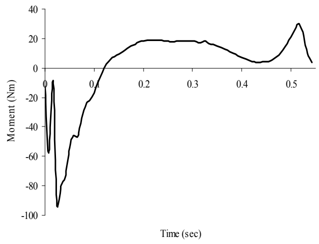


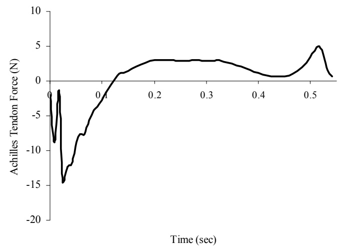


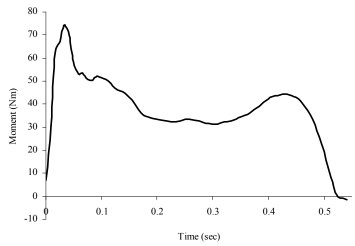


Participant 3

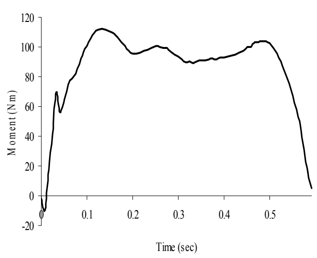
With heel insert

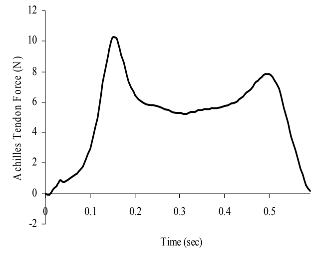


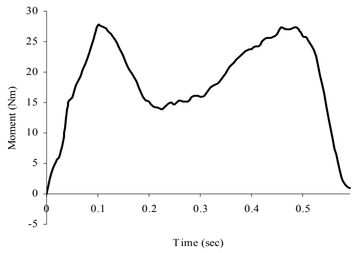




Without heel insert

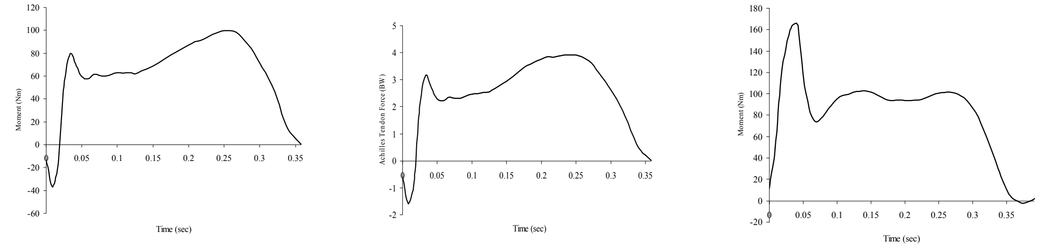




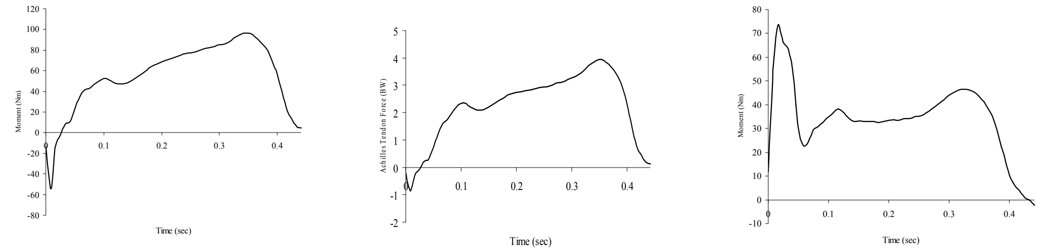


Participant 4

With heel insert

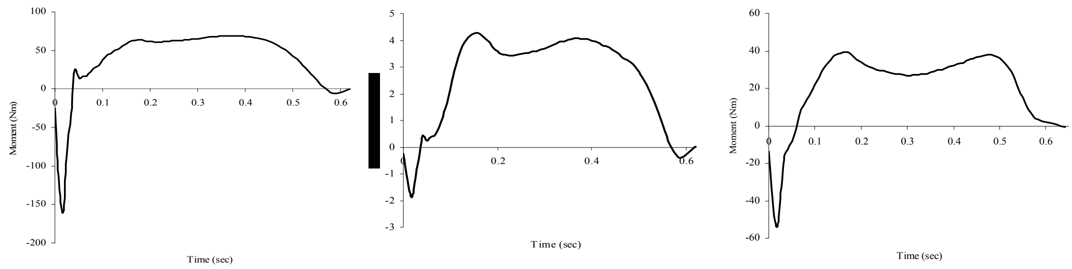


Without heel insert

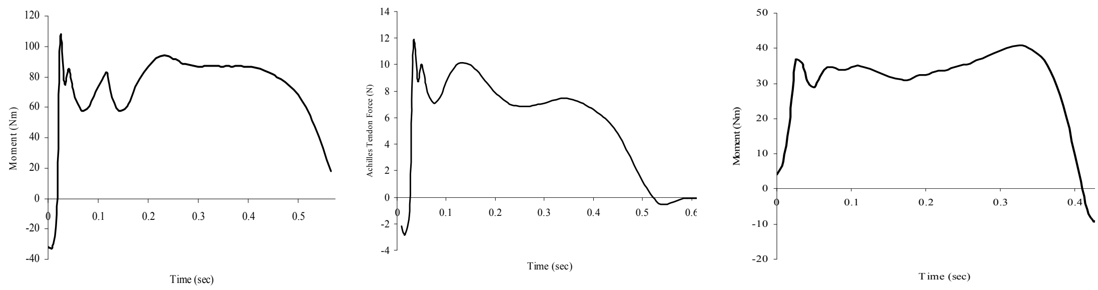


Participant 5

With heel insert

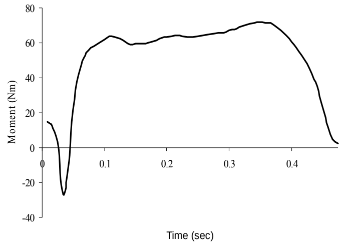


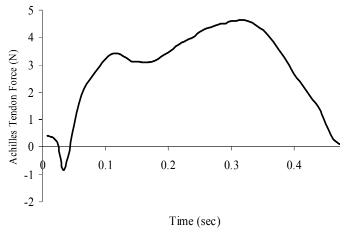
Without heel insert

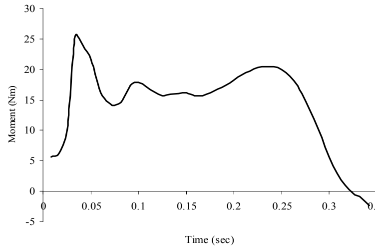


Participant 6

With heel insert







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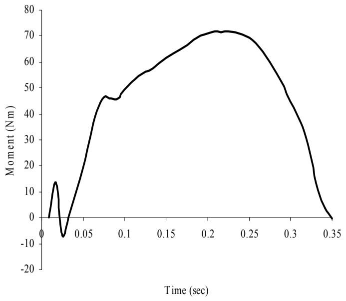
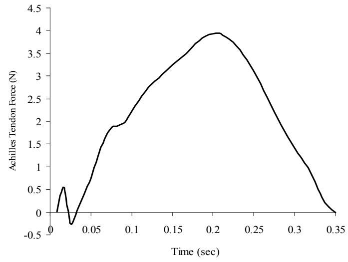
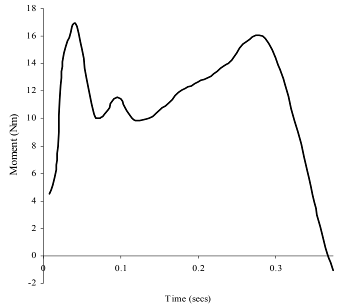
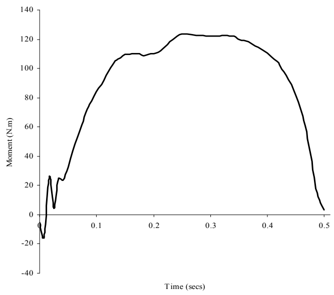


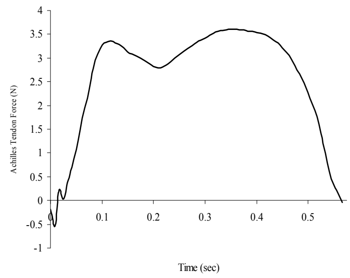


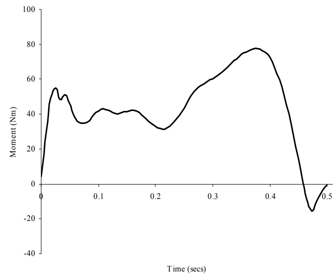


Participant 7

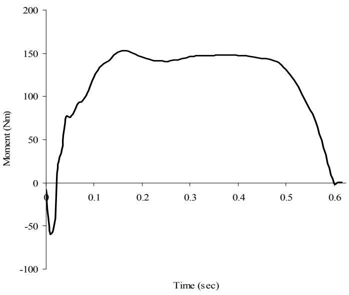
With heel insert

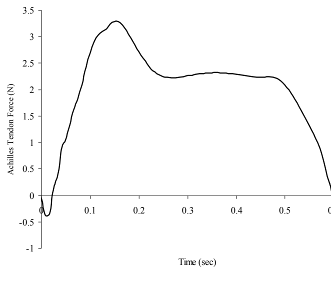


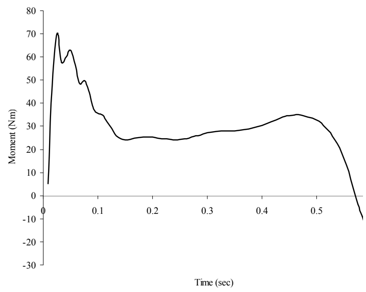




Without heel insert

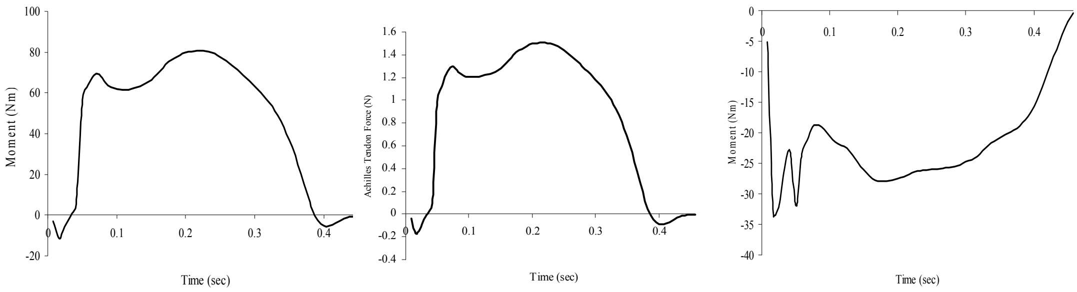




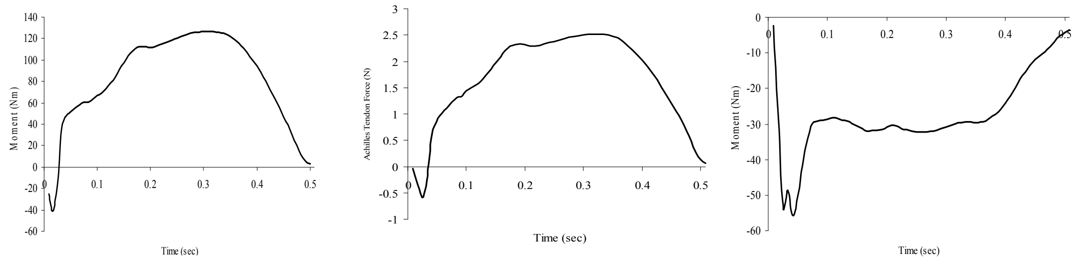


Participant 8

With heel insert

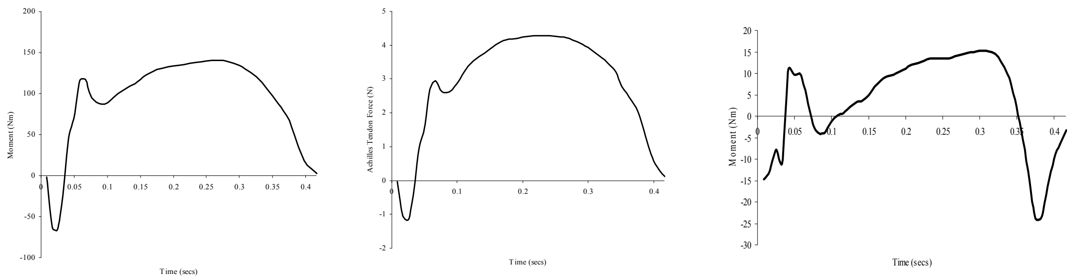


Without heel insert



Participant 9

With heel insert



Without heel insert

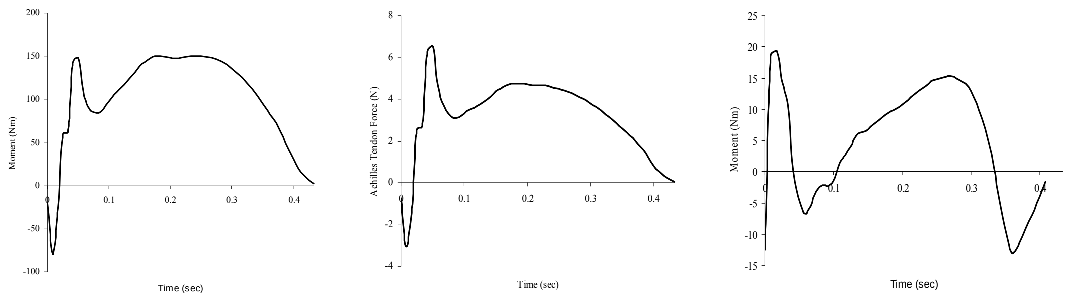


Figure 5.11. Samples of the dorsi-flexion and plantar flexion moment, Achilles tendon force and eversion moment time histories for each participant (Dorsi flexion- plantar flexion moment (Nm), Achilles Tendon Force (N), Eversion – Inversion moment (Nm) respectively)

The individual data were also statistically compared during the turning movement (Table 5.4). The paired samples t-tests showed that two participants experienced a significantly reduced peak Achilles tendon force (participants 3, 9). Four participants had a significantly different peak plantar flexion moment with the use of the heel insert (participants 2, 3, 4, 9), three of which were a reduction (participants 2, 3, 9). Three participants showed a significant reduction in the average plantar flexion moment loading rate (participants 2, 3, 9) and one participant showed a significant increase (participant 4). Three participants also showed significantly lower average Achilles tendon loading rate (participants 3, 4, 8).

Table 5.4 *Individual participant data whilst turning with and without a heel insert in a soccer boot for the measurements of peak Achilles tendon force (N) and (BW) and Plantar flexion moment (Nm) and the average loading of both (Nm.s)*

		Peak Achilles tendon force (N)	Peak Achilles tendon force (BW)	Average Achilles tendon force loading rate ($Nm.s^{-1}$)	Peak plantar flexion moment (Nm)	Average plantar flexion moment loading rate ($Nm.s^{-1}$)
1	Soccer boot	3664.1 (S.D. 1607.3)	4.3 (S.D. 1.9)	83667.2 (S.D. 55466.5)	107.9 (S.D. 12.7)	495.3 (S.D. 253.2)
1	Soccer boot + Heel insert	2914.3 (S.D. 1191.1)	3.4 (S.D. 1.4)	54425.5 (S.D. 15637.8)	105.8 (S.D. 34.7)	486.3 (S.D. 379.0)
		0.32	0.32	0.22	0.87	0.96
2	Soccer boot	2600.6 (S.D. 442.8)	3.6 (S.D. 0.6)	65897.1 (S.D. 33243.5)	105.6 (S.D. 4.6)	568.5 (S.D. 71.4)
2	Soccer boot + Heel insert	2332.1 (S.D. 469.3)	2.4 (S.D. 0.9)	68345.7 (S.D. 61666.6)	92.1 (S.D. 5.2)	472.1 (S.D. 122.5)
		0.23	0.23	0.96	**0.001*↓**	0.08
3	Soccer boot	7964.4 (S.D. 1289.8)	9.5 (S.D. 1.6)	43331.2 (S.D. 20552.7)	135.0 (S.D. 37.8)	974.4 (S.D. 492.1)
3	Soccer boot + Heel insert	1544.2 (S.D. 1409.9)	1.9 (S.D. 1.7)	3663.4 (S.D. 3022.3)	28.7 (S.D. 4.8)	66.0 (S.D. 13.4)
		0.001*↓	**0.001*↓**	**0.012*↓**	**0.03*↓**	**0.02*↓**
4	Soccer boot	2876.3 (S.D. 591.9)	3.9 (S.D. 0.8)	130742.7 (S.D. 57787.5)	103.3 (S.D. 2.6)	448.9 (S.D. 73.2)
4	Soccer boot + Heel insert	2675.4 (S.D. 734.1)	3.6 (S.D. 1.0)	83604.7 (S.D. 6924.2)	106.8 (S.D. 2.7)	344.5 (S.D. 13.9)
		0.51	0.51	**0.03*↓**	**0.001*↑**	**0.001*↓**

5	Soccer boot	5190.6 (S.D. 2621.9)	6.9 (S.D. 3.5)	187107.9 (S.D. 192348.5)	98.0 (S.D. 41.8)	2059.6 (S.D. 2791.0)
5	Soccer boot + Heel insert	3476.2 (S.D. 2532.9)	4.6 (S.D. 3.3)	68950.2 (S.D. 67966.5)	76.1 (S.D. 18.8)	612.6 (S.D. 797.7)
		0.15	0.15	0.09	0.19	0.17
6	Soccer boot	5561.2 (S.D. 4163.5)	6.7 (S.D. 5.1)	214980.0 (S.D. 226698.1)	74.7 (S.D. 8.1)	394.5 (S.D. 154.5)
6	Soccer boot + Heel insert	5766.1 (S.D. 2761.2)	7.0 (S.D. 3.4)	249604.4 (S.D. 222510.3)	68.7 (S.D. 8.3)	337.5 (S.D. 61.5)
		0.92	0.92	0.78	0.19	0.41
7	Soccer boot	2620.8 (S.D. 2735.3)	3.2 (S.D. 3.4)	214380.6 (S.D. 521319.8)	127.2 (S.D. 21.6)	564.6 (S.D. 167.9)
7	Soccer boot + Heel insert	1933.3 (S.D. 776.9)	1.6 (S.D. 0.9)	36331.2 (S.D. 14272.9)	139.1 (S.D. 39.8)	533.0 (S.D. 96.3)
		0.54	0.54	0.31	0.46	0.62
8	Soccer boot	3700.5 (S.D. 570.5)	4.3 (S.D. 0.7)	83069.5 (S.D. 18258.4)	148.1 (S.D. 12.2)	1054.4 (S.D. 1217.4)
8	Soccer boot + Heel insert	3393.5 (S.D. 727.9)	4.0 (S.D. 0.9)	64200.3 (S.D. 8091.7)	137.5 (S.D. 18.3)	981.9 (S.D. 962.4)
		0.39	0.39	**0.05***↓	0.23	0.91
9	Soccer boot	2635.4 (S.D. 744.5)	2.8 (S.D. 0.8)	54613.0 (S.D. 18597.1)	143.3 (S.D. 30.3)	704.4 (S.D. 214.3)
9	Soccer boot + Heel insert	1887.1 (S.D. 347.8)	2.0 (S.D. 0.4)	44638.4 (S.D. 18420.0)	110.3 (S.D. 19.3)	487.9 (S.D. 105.3)
		0.01*↓	0.24	**0.01***↓	**0.01***↓	**0.01***↓

↑ = increase with heel insert, ↓ with heel insert

Data are reported as mean and standard deviation (in brackets). Where, * denotes a significant difference at the $p<0.05$ level

The measurement of peak eversion moment was significantly different for three participants (participants 2, 7, 8) with participant 8 experiencing a significant increase whereas participants 2 and 7 exhibited a significant decrease. Three participants also experienced significantly different average eversion loading rates (Table 5.5). A reduction in average eversion moment loading rate was experienced by participant 2 and increased eversion moment loading rate was experienced by participants 3 and 4 when wearing the heel insert (Table 5.5).

Four of the participants experienced significantly different dorsi-flexion moments (participants 3, 6, 7, 8). Three of these were significantly reduced with the heel insert (participants 6, 7, 8) and one showed significantly greater peak dorsi-flexion moment (participant 3) (Table 5.5). Three of the participants exhibited a significant change in average dorsi-flexion moment loading rate, one of which showed a significant decrease (participant 3) and the others showed a significant increase (participants 4, 8 [Table 5.5]).

Table 5.5 *Individual participant data whilst turning with and without a heel insert in a soccer boot for peak eversion and dorsi-flexion moment (Nm) and the average loading rate of both (Nm.s)*

		Peak eversion moment (Nm.)	Average eversion loading rate (Nm.s^{-1})	Peak dorsi flexion moment (Nm)	Average dorsi-flexion loading rate (Nm.s^{-1})
1	Soccer boot	54.9 (S.D. 13.9)	416.5 (S.D. 324.4)	-62.3 (S.D. 25.2)	-6491.7 (S.D. 3976.5)
1	Soccer boot + Heel insert	51.3 (S.D. 6.4)	317.3 (S.D. 412.3)	-66.6 (S.D. 59.2)	-5212.7 (S.D. 3199.5)
		0.52	0.59	0.85	0.51
2	Soccer boot	31.9 (S.D. 24.0)	1977.0 (S.D. 1168.9)	-82.1 (S.D. 54.9)	-8068.0 (S.D. 7312.1)
2	Soccer boot + Heel insert	5.2 (S.D. 13.3)	46.6 (S.D. 100.3)	-60.8 (S.D. 0.1)	-4950.4 (S.D. 3301.8)
		0.03*↓	**0.02***↓	0.34	0.32
3	Soccer boot	29.8 (S.D. 12.6)	293.1 (S.D. 169.5)	-14.1 (S.D. 7.6)	-1230.1 (S.D. 125.1)
3	Soccer boot + Heel insert	59.8 (S.D. 59.4)	4081.1 (S.D. 4634.3)	-128.3 (S.D. 45.2)	-7029.5 (S.D. 5221.2)
		0.29	**0.03***↑	**0.001***↑	**0.004***↑
4	Soccer boot	31.6 (S.D. 9.3)	109.6 (S.D. 17.6)	-17.8 (S.D. 4.9)	-905.6 (S.D. 216.7)
4	Soccer boot + Heel insert	32.8 (S.D. 9.4)	195.8 (S.D. 37.9)	-16.3 (S.D. 3.4)	-666.5 (S.D. 88.9)
		0.76	**0.001***↑	0.461	**0.007***↓
5	Soccer boot	39.3 (S.D. 9.2)	706.7 (S.D. 634.9)	-31.7 (S.D. 8.5)	-2253.0 (S.D. 1257.2)
5	Soccer boot + Heel insert	38.0 (S.D. 8.8)	441.3 (S.D. 638.7)	-43.6 (S.D. 29.6)	-3443.6 (S.D. 3377.4)
		0.77	0.40	0.31	0.4
6	Soccer boot	23.3 (S.D. 12.2)	757.3 (S.D. 484.1)	-20.5 (S.D. 11.7)	-1096.3 (S.D. 750.9)
6	Soccer boot + Heel insert	16.3 (S.D. 8.4)	546.0 (S.D. 402.2)	-7.8 (S.D. 7.6)	-440.6 (S.D. 478.3)
		0.25	0.40	**0.05***↓	0.15
7	Soccer boot	122.7 (S.D. 24.3)	613.1 (S.D. 414.7)	-76.2 (S.D. 48.7)	-7683.3 (S.D. 6141.0)
7	Soccer boot + Heel insert	85.5 (S.D. 31.8)	1486.2 (S.D. 2573.7)	-35.8 (S.D. 23.7)	-3724.6 (S.D. 3154.3)
		0.02*↓	0.14	**0.03***↓	0.1

8	Soccer boot	16.6 (S.D. 8.9)	170.7 (S.D. 208.5)	-70.7 (S.D. 21.5)	-8193.5 (S.D. 3197.3)
8	Soccer boot + Heel insert	33.9 (S.D. 19.6)	674.10 (S.D. 786.64)	-38.4 (S.D. 28.5)	-3561.4 (S.D. 3139.0)
		0.05*↑	0.23	**0.05*↓**	**0.03*↓**
9	Soccer boot			-32.8 (S.D. 33.5)	-3142.4 (S.D. 4045.8)
9	Soccer boot + Heel insert			-19.6 (S.D. 19.3)	-1767.3 (S.D. 2272.1)
				0.34	0.41

↑ = increase with heel insert, ↓ with heel insert

Data are reported as mean and standard deviation (in brackets). Where, * denotes a significant difference at the $p<0.05$ level

5.4. Discussion

Having shown that the group analysis of estimations of Achilles tendon or lateral ankle loading were not significantly different, yet individual differences were significant, these findings have implications on future recommendations for the use of heel inserts in soccer. These implications as well as suggested reasons for the results are discussed below.

5.4.1. Achilles tendon and ankle loading

The moment-time history during running showed that participants exhibited a small initial dorsi- flexion moment, which changed into a larger plantar flexion moment which peaked during mid-stance. This general pattern is in agreement with the plantar flexion moment time-history presented by previous authors (Reinschmidt & Nigg 1995; Winter, 1983; Scott & Winter, 1990). The observation that some participants did not show a dorsi-flexion moment is also consistent with a previous study (Reinschmidt & Nigg 1995). The average magnitude of the plantar flexion moment recorded was 196.9 Nm (S.D. 73.49) and 192.5 Nm (S.D. 53.25) for the no heel insert and 10 mm insert conditions respectively. These values are at least 20% smaller than those presented by Reinschmidt & Nigg, (1995), but are similar to those of Winter (1983) and Scott and Winter (1990). The reason for the difference between the values given by Reinschmidt & Nigg (1995) and those presented in the current investigation is likely to be related to the lower running speed chosen in the current study (4.6 m/s vs. 3.81 m/s, respectively).

At least 85% of the force during dorsi flexion is attributed to the work of the triceps surae muscle group and thus the Achilles tendon (Scoot & Winter, 1990). Using plantar flexion moment to estimate the forces during dorsi-flexion, the comparison of with and without heel inserts yielded data that were not significantly different between conditions. This is consistent with an earlier study by Reinschmidt & Nigg (1995), but indicates that the proposed methodological changes, specifically the use of three-dimensional data, failed to improve the sensitivity of this measurement. One proposed explanation for this finding is that the response to the heel insert intervention is highly individual (Dixon & Kerwin, 1998; 2002; Reinschmidt & Nigg 1995). The paired samples t-tests performed for each participant showed that only two of the nine participants exhibited significant differences in peak plantar flexion moment. One of these participants exhibited significantly greater peak moment and the other a significant decrease in the heel insert condition. This supports the idea that the response to heel insert conditions are highly individual, although for most there did not seem to be a significant affect.

The estimation of Achilles tendon force provided values of approximately 5 and 6 times the body weight of the participants, which is within the magnitudes reported by Komi (1990) (up to 12 times body weight) when in vivo Achilles tendon forces were measured. Values are also similar to those reported by Dixon and Kerwin (1998, 2002) when using an inverse dynamics approach.

Consistent with the findings of Dixon and Kerwin (2002), the peak Achilles tendon force measured in the current investigation was not significantly different between the heel insert and non-heel insert conditions for the group mean data. As two participants showed significant reductions in peak Achilles tendon forces with the heel insert, there is potential in these participants that the risk of injury may be reduced. In contrast, as one participant showed a significant increase in Achilles tendon force, this suggests that the risk of injury increases in some participants with the inclusion of the heel insert. The observation of an increased Achilles tendon force is consistent with the findings of Dixon and Kerwin (1998) and emphasises further the individual response that can occur to the heel insert.

Achilles tendon magnitudes were specifically lower with a heel insert for participants 1 and 8 during running. Plantar flexion moment however, was not significantly different.

An inference can therefore be made that the change in Achilles tendon force is expected to be in response to a change in the moment arm length, resulting from a change in the position of the heel relative to the forefoot. In this situation the peak moment arm length remains longer, which would reduce the force applied to the tendon. This change in foot position is therefore consistent with the mechanism of Achilles tendon injury put forward in past explanations (Clement et al., 1984; Leach et al., 1981).

Performing different movement patterns also failed to reveal group differences, which is in contrast to those studies of playing surfaces and footwear (Dixon & Stiles, 2003; Queen et al., 2008), yet significant individual differences were evident. During the turning movement, the dorsi-flexion and plantar flexion moment-time histories for each participant were similar to those presented by Morlock and Nigg (1991). The participants in both studies showed an increase in the negative moment that occurs in response to increased plantar flexion. Throughout the mid-stance of the turn, the moment changes direction in response to increased dorsi-flexion. The moment-time history presented by Morlock and Nigg (1991) remained negative suggesting that a dorsi flexion moment was still being applied, albeit to a lesser extent than that experienced earlier in the stance. In contrast, participants in the current study exhibited a positive moment. This difference may be explained by appreciating the differences in the movements performed. Although similar, Morlock and Nigg (1991) used a side shuffle rather than a turn, thus the range of motion may have been less than the turn employed in the current investigation.

The variance between subjects was larger during turning than for running. This may be due to the individual nature of the turning movement and the self selected speed of movement. Most however, had similar patterns. Participants 3 and 9 showed a significantly reduced plantar flexion and Achilles tendon force. It is therefore likely that both a reduced moment and lengthened moment arm are contributing to the experience of significantly reduced force. Conversely, in the participants who did show a reduction in plantar flexion moment but not Achilles tendon force, it is suggested that the moment arm in these individuals may not have changed sufficiently for differences to be observed. This again indicates the individual response to the heel insert intervention. The greater number of participants responding to the heel lift during this movement compared to running also supports the rationale for using additional movements when assessing the different interventions.

The rate of loading measurement has been suggested to be a more important characteristic in the aetiology of overuse injury (Lees & McCullagh, 1984; Radin et al. 1991) and has been able to distinguish differences between with and without heel insert conditions when the measurement of peak force cannot (Dixon & Kerwin, 2002). Dixon and Kerwin (2002) found that the measurement of average loading was able to differentiate between footwear conditions with and without a heel insert. They found that significantly lower average rates of loading were observable when running in a 15 mm heel insert compared to when no heel insert was used. However, the average loading rate with a 7.5 mm insert was not significantly different. In the current investigation the use of a 10 mm heel insert was expected to reduce the average Achilles tendon loading rate, replicating the findings of Dixon and Kerwin (2002) when using a 15 mm heel insert. Having observed that during both running and turning, the calculated average loading rate of Achilles tendon force and plantar flexion moment were not significantly different between the footwear conditions, evidence is provided to suggest that in order to significantly reduce average Achilles tendon loading rate in a group of participants, small heel inserts (<10 mm) are not sufficient.

Collectively, these findings do not further the understanding of the mechanism behind the reduction of Achilles tendon injury with the use of a heel insert, although the data does support the idea that the response to the heel insert is individual. However, the effect of the heel insert may change if the surface was different, since previous biomechanical investigations have shown that the response of different footwear conditions is influenced by the surface on which the footwear is tested (Dixon et al., 2008; Chapter 3).

Reinschmidt and Nigg (1995) suggested that the aetiology of Achilles tendon injury is not related to the peak forces that occur within the tendon. Instead, they have suggested that it is possible that the source of inflammation may be related to calcaneal friction. This is related to a change in the relative position of the calcaneous with respect to tibia. Therefore, by using the heel insert, the angle may be less variable throughout the stance phase, and the tendon may be less irritated. For example, the size of this angle was said to relate to inflammation of the distal part of Achilles tendon, especially when people have ‘predisposed’ anatomy and/or orientation of the calcaneous when wearing shoes with low heel lift. This may cause the Achilles tendon to rub on the superior calcaneous tuberocity, leading to chronic tendon inflammation which can be aggravated in runners with large eversion. Reinschmidt & Nigg (1995) suggested that this aetiology may only apply to insertional Achilles tendonopthy. As insertional Achillies tendon pathology occurs commonly in sports that involve running such as soccer (Nunley, 2008) and that this contributes to the 94% of Achilles tendon injuries that were either tendonitis or paratendonitis, this mechanism may play a role in the reduction of Achilles tendon injury and thus may explain why no differences were experienced for the measurements taken in the current investigation.

Another problem that arises from the use of peak plantar flexion moment and Achilles tendon force is that the magnitude only refers to axial loads or a stretching of the Achilles tendon (Reinschmidt & Nigg, 1995). The change in heel height may change non-axial loads, such as shear and/or bending. This may be affected by the amount of pronation that occurs which has been shown to be influenced by heel height (Bates et al., 1978; Stacoff & Kalin, 1983). Also presented in Chapter 4 was evidence to suggest that peak rearfoot eversion did not differ significantly between footwear conditions when different heel inserts were tested. Despite this finding, the individual nature of the heel lift response may have concealed such a change in most participants. In particular, those participants with greater eversion magnitude may observe greater benefit from the heel insert than those participants with lower magnitudes.

Reinschmidt and Nigg (1995) expected that the success of heel insert interventions results from a combination of a reduction of both calcaneal friction and non-axial loading. Likewise, Reinschmidt and Nigg (1995) highlighted that the individual response in the magnitude of peak moment may contribute alongside these factors to explain the success of the treatment. Dixon & Kerwin (1999) also found that the peak strain generated was significantly greater without a heel insert, suggesting that this may also influence the success of the intervention.

What the evidence does suggest in that neither the use of the three-dimensional data nor the use of shod running improve the sensitivity of joint moment or Achilles tendon force calculation to such an extent that significant differences could be observed. Even when the inclusion of the heel insert was tested in a soccer boot, no significant differences were detected. Because of these results, it is suggested that the heel insert does not significantly reduce the forces for the sample of participants used. However, the observation that some individuals experience a significant reduction in peak force and average loading during running and turning and that one participant experienced significantly greater peak Achilles force when running, highlights that the heel insert may serve to increase the risk of injury. As such, the use of heel inserts for treating overuse Achilles tendon complaints as well as other injuries, such as heel pain, may be successful for some participants, yet in a select few, injury risk may be greater. This may related to anatomy characteristics that predispose the participants to injury. Thus this intervention should be used with caution and examined on an individual basis.

5.4.2. Inversion injury risk

Reinschmidt and Nigg (1995) found that during running the magnitude and time of occurrence of the initial dorsi-flexion moment were significantly affected by heel height. Likewise, peak plantar flexion movement was significantly greater when turning with the heel insert placed into the soccer boot in the previous chapter. Despite this, it was found that no significant differences were observed for maximum dorsi-flexion moments or average loading rate; thus this would seem not to agree with the previous evidence.

In the current investigation, peak eversion moments were also measured to represent the internal forces occurring in response to rearfoot inversion (Morlock & Nigg, 1991; Park et al., 2005). The shape of the eversion-inversion moment time history was similar to those presented by Morlock and Nigg (1991). The magnitude of the peak eversion moment also replicated those values presented by Morlock and Nigg (1991) during a side-step task in different tennis shoes. Likewise, the values were also similar to those presented by Park et al. (2005) for a “V” shape cutting manoeuvre. These authors measured the peak eversion moments when participants wore different soccer boots with contrasting stud designs to represent the moment occurring during an inversion movement. Magnitudes of 50-54 Nm were experienced which were similar to those found in the current investigation, although Park et al. (2005) used a running speed of 4.0 m/s and a cutting movement was performed, which is not akin to the turning movements used in the current study.

Eversion moments were not different when comparing the control to the heel insert condition, yet single subject analysis did show that for three participants there was a significant increase in either peak eversion moment or average eversion moment loading rate. It is therefore, likely that these participants are at a greater risk of lateral ankle damage. This heel insert may therefore further contribute to the high incidence of ankle damage already present in soccer (Woods et al., 2003). Conversely, it appears that for another participant there was a significant protection with the use of the 10 mm heel insert. If during a turning movement the magnitude and rate of lateral forces are increased through increased surface traction or movement speed, this may amplify the effect of any instability due to the heel insert so that participants could surpass the thresholds of injury in game situations.

A limitation of the current investigation is that the calculated moments occurring during turning do not accurately represent the forces occurring within the specific ligaments. The forces of these ligaments contribute to the magnitude of eversion and dorsi-flexion moments but other ligaments and musculature play some part in the restriction of the ankle joint. This may make the observation of significant differences difficult.

The lack of significant differences may also result from the type of turning task performed. The choice of movement was a complete turn, that aimed to represent a change in direction occurring in soccer, where the risk of maximum inversion occurring is greatest. This presumption was made based on anecdotal and personal experience. Instead, it may be that during other types of turning such as side stepping and cutting, more severe inversion could be observed, indicating a greater risk of injury. Likewise, the participants performed turning movements that were at a self selected speed and this may have been well within their comfort levels. As a consequence, the difference in inversion angle may not be as great as if the participants were asked to run and turn at greater speeds. The lack of significant differences may also relate to the nature of acute injury. Traumatic or acute injuries often occur suddenly in response to an unexpected change in foot angle resulting from an unanticipated movement or surface condition. As turning is an unaccustomed movement compared to running and that participants were aware of the timing of the turn along with the surface being uniform, this may have prevented significant differences in maximum inversion and dorsi-flexion moment from being observed. Further still, if the turn was performed on a different surface or the uniformity of the rubber particles was more inconsistent, this may predispose the foot to even greater instability. Finally, the use of an alternative surface-footwear combination may also influence the traction. Consideration of the surface conditions when testing footwear have been advocated in the previous chapters as well as by Dixon and colleagues (2008). Future research should therefore focus on the effect of changing turning speed and knowledge of when the turning movement will occur on the magnitude of lateral ankle loading when using heel inserts. Similar considerations should also be given to the footwear worn and surface consistency and cushioning provided. However, there may be some difficulty obtaining ethical approval for such investigations as these are high risk movements.

In summary, the evidence presented indicates that at first glance the risk of inversion injury did not appear to change with and without the heel insert when group mean comparisons are made. Despite this, when individual data were analysed some significant differences were shown. In such cases the use of the heel insert is not appropriate. Changes to the boundary conditions and the turning speed may also increase the moments produced so that injury thresholds are reached. As such caution is needed when using heel inserts in soccer boots.

5.4.3. Conclusion

In conclusion, despite the reduction and treatment of Achilles tendon injury reported in previous literature with the use of a heel insert, the group means for the measurement of peak plantar flexion moments or average plantar flexion loading rate were not significantly different when running or turning whilst wearing a heel insert. These results do not support the original hypothesis that plantar-flexion moments and their rate of loading would be reduced with a heel insert. Likewise, neither peak Achilles tendon force nor average loading rate were found to be significantly reduced with the heel insert. This suggests that these measurements were unable to distinguish between heel insert conditions for a group of participants. Despite these results, single subject analysis suggests that some participants did exhibit significant individual differences, highlighting possible mechanisms behind a reduction of injury risk in some players. In one participant, however, significantly greater peak Achilles tendon forces were shown, indicating an increased risk of injury.

The lack of significant group difference in peak eversion or dorsi-flexion moments does not support the study hypotheses that these variables would be greater with the heel insert condition. For some participants the addition of the heel insert was speculated to increase the risk of ankle sprain through either an increase in peak eversion or dorsi-flexion moments or the average loading rate of these measurements. Therefore, these findings suggest that the use of the heel insert in sports involving dynamic movements should be performed with care and that the use of the heel insert may require consideration of the individual player characteristics. This may require knowledge of previous injury, which is related to ankle ligamentous stability (Tyler et al., 2006; Chomiak, et al., 2000; Ekstrand & Gillquist, 1983; Woods et al., 2002), as well as other risk factors associated with ankle sprains such as muscular imbalances and proprioception problems linked to the gait mechanics (Willems et al., 2005; Willems et al., 2002). Participants with such characteristics may then be recommended to avoid the use on the heel insert particularly during lateral movements.

Chapter 6
Conclusions, Recommendations and Future Directions

6.1. Thesis background

The playing surface and footwear have been cited as risk factors that contribute to the high risk of injury in soccer (Woods et al., 2002; Hawkins et al., 2001). Soccer is traditionally played on a natural turf surface (Ekstrand et al., 2006), which becomes hard during the preseason, and which has been linked to the increased risk of injury in this sport particularly during preseason and early in-season (Woods et al., 2003; Woods et al., 2002; Hawkins et al., 2001). However, third generation artificial turf surfaces have also been developed for use in soccer (Ekstrand et al., 2006). These surfaces are purported to provide an alternative to natural turf surfaces and are designed to remain consistent throughout the year, which may help reduce the risk of injury in soccer.

Understanding of the biomechanical response of soccer players on different playing surfaces is somewhat limited due to the difficulty in incorporating natural and artificial turf into a biomechanics laboratory (Stiles et al., 2008), and the inflexibility of using biomechanical equipment outdoors. Likewise, similar limitations have existed when comparing different soccer footwear. As such, the main objective of this thesis was to use biomechanical techniques to answer the question *"what effect do the playing surface and footwear variables have on the biomechanical response of soccer players"*.

6.2. Main study findings, implications and contribution to biomechanical research

To examine this question three experimental chapters are presented. The first study presented in this thesis (chapter 3) specifically compared the biomechanical response of soccer players on natural and third generation artificial surfaces at different times of the year using variables which have been related to acute and overuse injury. This study therefore aimed to investigate how the biomechanical variables can differ between surfaces, whilst understanding whether a seasonal effect is present when comparisons are made. A second aim was to understand the effect that soccer footwear has on the biomechanics of soccer players, as well as whether the response to the footwear is specific to the surfaces on which the movement is performed. Lastly, the study aimed to investigate the role that changes in environmental conditions can have on the same natural turf and third generation surface in influencing the biomechanical response of the participant. In-shoe pressure technology was utilised in this study, alongside kinematic observations obtained from VHS cameras, to assess soccer players during running and turning movements. A protocol was developed in the first study to enable the collection of biomechanical data during these movements. Analysis of players was performed on council owned natural and third generation artificial soccer surfaces at two times of the year when contrasting environmental conditions were found.

The study showed that some measurements associated with lower extremity loading were different between natural and third generation surfaces. These measurements included peak impact force and medial heel and peak pressure loading rate at the lateral heel, and peak loading rate and peak pressure at the first metatarsal. However, these differences depended upon the time of year the surfaces were compared and the movements that were performed. For example, the peak pressure at the medial heel and peak pressure loading rate at the lateral heel were significantly greater when turning in March on the third generation and natural turf respectively. On the other hand, peak impact forces, peak pressure at the first metatarsal whilst turning and peak rate of loading when running were significantly greater in May on the natural turf. These results indicate that the use of third generation artificial surfaces and natural turf can influence the loading of the lower extremity, but also highlights the importance of selecting an appropriate time of year, environmental conditions and movements when assessing the characteristics of different playing surfaces in future investigations as this may influence the conclusions drawn.

This investigation also measured the biomechanical changes that occurred when the participants ran and turned on the same surface at the two periods of the year that were analysed. This allowed evaluation regarding how the environmental conditions influence the playing surface and determine the biomechanical response of the soccer player. It was observed that during both running and turning, significantly greater peak pressures and peak pressure loading rates were shown in May when warmer and drier conditions were reported compared with the same surface in March. This was evident for both natural and third generation artificial surface. This shows that as was expected, the natural turf became hard and possessed less cushioning when warmer and drier conditions were evident in May. However, the change in pressure and pressure loading rate on the third generation surface was in contrast to the belief that the artificial turf would remain consistent between sessions. This may have resulted from the change in the environmental conditions, but it may also relate to the amount of use these public playing surfaces receive. This would highlight the possible need for improved maintenance of this council owned facility.

Because of the findings presented in study 1, it is recommended that by making hard natural turf surfaces more cushioned through increased watering, some of the biomechanical measurements associated with injury risk may be reduced. However, this can be expensive and the use of watering is a pertinent issue for those people who are environmentally aware. Alternatively, the risk of injury may reduce if the soccer player uses a third generation artificial surface, and this may be particularly useful during preseason and early in-season when natural turf surfaces are described as mainly dry and hard, and the risk of injury is disproportionately high. However, in professional stadia, these surfaces can not be interchanged at will. Therefore, there may have to be a compromise between using the natural turf in matches and third generation artificial surface in training. This may pose another problem as the switch from one surface to another may prevent a surface specific muscular hypotrophy in some participants, thus predisposing them to possible injury.

Although the investigation presented in Chapter 3 (study 1) advocates the use of a third generation artificial surface, the construction of the shock pad density was also highlighted as a potential factor that could influence the cushioning provided by third generation surfaces (McNitt et al., 2004; Fleming et al., 2008). Mechanical tests have been performed to understand how the cushioning provided by the shock pad density can change. However, these tests do not faithfully replicate the behaviour of the participant and provide no biomechanical data to indicate how the shock pad influences the performer. Therefore, chapter 4 presents the findings from a comparison of soccer players during running and turning on different shock pad densities using external biomechanical variables. By including and testing a third generation turf in a biomechanics laboratory, this set-up provides novel insight into the effect of the shock pad density on the biomechanics of a soccer player. This contributes to the understanding of third generation artificial turf surfaces on the performer.

The results of study 2 indicated that there was a significantly lower peak pressure at the first metatarsal and lower peak impact force on the more cushioned shock pad when turning. However, significantly reduced impact force was only observed when measuring this variable with in-shoe pressure insole technology, and not the force plate.

As a consequence the use of more cushioned shock pads may influence the risk of injury. It is therefore recommended that the shock pad cushioning is an important consideration when designing third generation artificial surfaces. The results also highlight the improved sensitivity and ability of the in-shoe pressure technology to detect differences between differently cushioned surfaces, when compared to the traditional force plate. Consequently, insole pressure technology should be used by researchers when comparing footwear and surfaces in future investigations.

In both study 2 and study 3 (chapter 5), investigations were performed into how additional material placed into the boot can influence the soccer players' biomechanical characteristics associated with injury risk. Study 2 aimed to investigate the biomechanical mechanisms behind the successful application of heel inserts (10 mm Sorbothane heel insert) and soccer specific insoles (Sorbothane) in reducing injury risk. These devices were tested using external biomechanical measurements associated with injury. No evidence was provided to support suggestions that the cushioning insole would reduce lower limb loading during running or turning. Whilst this suggests that this soccer-specific insole may not be beneficial, the potential of the insole to reduce loading if used on a harder playing surface cannot be dismissed. The observation that the heel insert resulted in a significant increase in the time to peak impact force in turning, whilst there were no changes observed in peak ankle dorsi-flexion, suggests that this intervention is successful in treatment/prevention of Achilles tendon injury through increased impact attenuation, rather than through a lowering of the strain on the Achilles tendon. However, as this was not shown during running, the mechanisms behind the success of heel inserts during this task were unclear.

Study 2 also aimed to understand the effect that the inclusion of the insert would have on the biomechanical behaviour during lateral movements. The observation of a significantly greater plantar flexion with the heel insert when turning, suggests greater strain would be placed on the ATFL, yet the inversion movement, most commonly associated with these injuries, was not significantly different. This may have resulted due to the insensitivity of this external measurement to the change in footwear condition.

In the final experimental chapter (Study 3), it was suspected that internal force measurements may be more revealing than external measurements used in chapter 4 and that by using this methodology, the mechanism behind Achilles tendon injury reduction may be clearer. However, despite the use of shod movements and three-dimensional moments and moment arms, there were no significant differences for any of the measurements associated with Achilles tendon damage when either running or turning. Joint moments were also used to assess the risk of inversion injury, but again, no significant differences were observed for these measures using group mean comparisons.

In contrast, the responses to the heel insert were found to be highly individual. Individual data analysis showed that some participants experienced significantly lower peak plantar flexion moment, Achilles tendon force or a reduced average loading rate of these measurements with the heel insert intervention. However, it was also shown that for one participant, peak Achilles tendon forces actually increased with the use of the heel insert. Because of these findings, the mechanism behind the success of heel inserts remains unclear, but may indicate that the success is multi-factorial and thus is highly individual. Likewise, the risk of inversion injury was also specific to the individual. Two participants showed an increased risk of injury and because of this it is recommended that the use of heel inserts in soccer should be performed with care. The magnitude of this load is likely to further increase, as the speed of the movement, and the traction between the surface and footwear, is increased. Therefore, if the heel insert is used, it is advised that performers returning from an ankle injury should avoid lateral movements until strength is fully restored. Care must also be taken as the results of the insert study were collected on a third generation turf with participants wearing moulded boots. The response of the participants may have been different if the footwear or surfaces were changed. If harder surfaces and longer studs were worn, this may create greater traction and may make control of the rearfoot movement more difficult. This could then lead to an overloading of the peroneal musculature and increase the risk of injury to the ankle ligaments. The same observation may also occur if participants performed greater movement velocities. In addition, greater risk of injury may occur when participants are performing other cutting movements and when the movement is unanticipated. Consideration of all of these factors is required before the use of heel inserts should be recommended for use in soccer.

In summary, the important findings uncovered by the research of playing surfaces and footwear in soccer within this thesis are that natural and third generation playing surfaces can be differentiated by using biomechanical measurements associated with lower extremity loading. The shock pad density can also influence the loading of the soccer player particularly during turning. Individual data is also important when uncovering possible reasons behind the successful prevention of injury with devices such as heel inserts and also the potential risk of sustaining injury. The data presented also showed that the time of year, movements performed and the measurements used determine the differences that can be observed between the surface and footwear conditions. Additionally, the surface conditions further influence the observation of biomechanical differences between the playing surfaces.

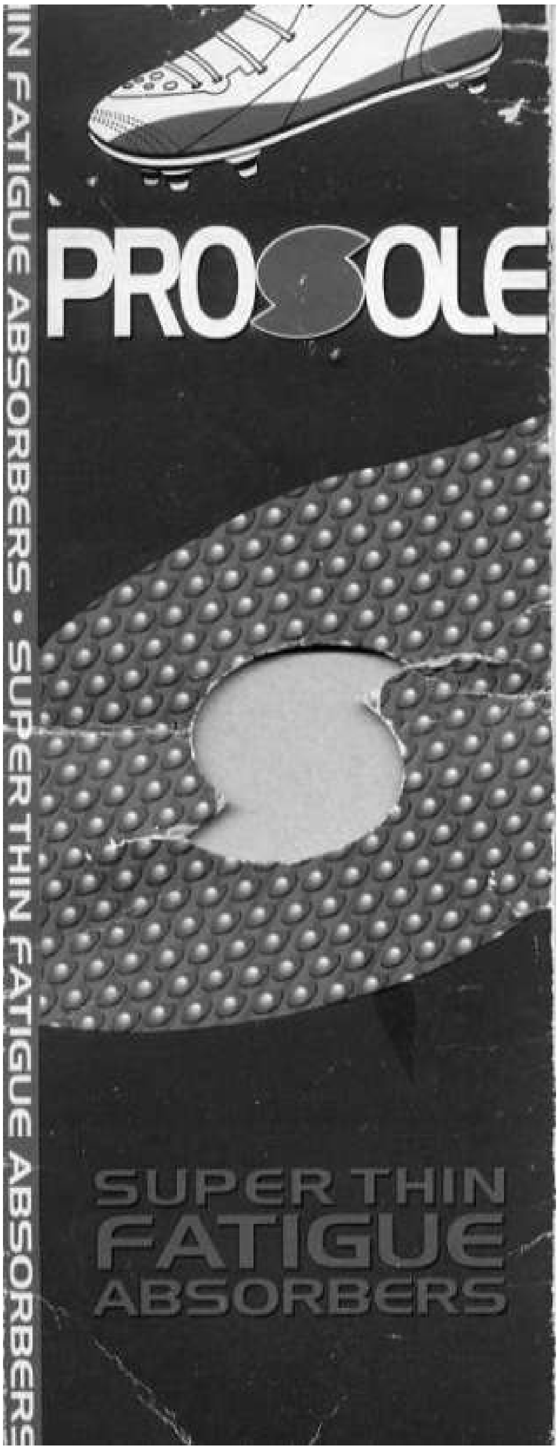
6.3 Future research

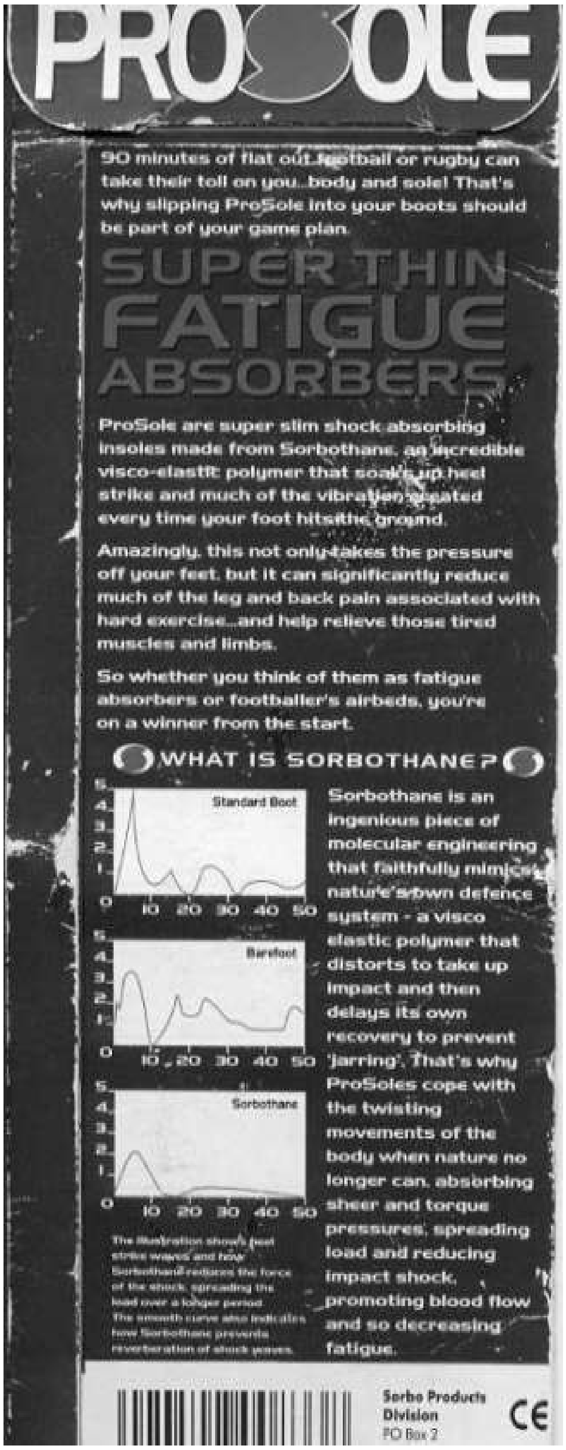
To further understand the response of the soccer player to changing playing surface and footwear, it may be important to appreciate the role of internal characteristics such as the natural ankle dorsi-flexion or subtalar range of motion that a participant exhibits. Likewise, it would be useful to test participants who have had previous injuries in order to assess whether differences could be observed in these participant groups and whether the interventions can prevent reoccurrence. Injuries in soccer are also common when players are fatigued. In particular, overuse Achilles tendon injury tends to result from eccentric loading of fatigued muscles (Clement et al., 1984). Therefore, the mechanism behind the effectiveness of the heel insert may be more apparent in fatigued individuals. As most biomechanical research tests the effect of different footwear and surfaces on un-fatigued individuals, this limitation requires further investigation. Further still, measurements of acute injury risk in studies of this type are limited to sub-maximal intensity movements in order to prevent injury during participation. This may allow structural control from the muscles to resist excessive lateral movement and therefore significant differences are hard to observe. Instead, if greater speeds are used, more extreme movements may occur and significant differences could be more likely. However, this may put participants at a greater risk of injury and additional care and consideration would be needed for this research to be performed. Future work could also investigate how a change to the playing surface influences the response to a heel insert and cushioning insoles intervention. As biomechanical responses change with different surfaces (Dixon et al., 2008), a cushioning insole or heel insert may be more effective on a hard natural surface compared to those used in the current investigations.

Conclusions based on Study 1 regarding footwear were limited to the footwear used. The participants' biomechanical response to footwear has been shown to differ across research publications (Queen et al., 2008; Dixon et al., 2008). This is somewhat related to the brand and model used, as each has slightly different plantar sole, mid-sole and upper boot construction, despite the description of the boot in these studies being similar (i.e. screw-in). Therefore, if a more diverse range of footwear, possibly including boots described as bladed, were used, more diverse biomechanical differences related to both acute and overuse injury may be observable. Study 1 also showed that despite similarities in the mechanical cushioning properties of the surface, significant biomechanical differences were shown. This highlights the inability for mechanical tests to fully indicate how the participant responds to the surface. Therefore, future biomechanical research of footwear-surface traction and cushioning can be used alongside engineering approaches to modify mechanical tests so that they react in a more similar manner to that of a human. This would then provide a mechanical method by which large number of surfaces and/or types of footwear can be tested to replicate the behaviour of the participant more closely.

By collecting the data presented in this thesis, different limitations of previous investigations have been addressed. This ensures that each study contributes something new to the understanding of the biomechanical effect of playing surfaces and footwear in soccer. These novel aspects of the studies provide interesting and more accurate insight in to the behaviour patterns of soccer players when surface and footwear variables are manipulated. One important insight gained relates to the vital consideration one must give to the time of year different surfaces are tested, as the environmental conditions can impact on the conclusions drawn from these comparisons. Thus, these findings must be utilised in future biomechanical studies. Likewise, the findings clearly demonstrate that whilst footwear change may reduce the risk of one injury, the risk of other injuries may increase. Thus, similar investigations should also appreciate this effect. The evidence also champions the use of in-sole data to detect differences between playing surfaces. Finally, it does appear that footwear and playing surface variations do have the ability to influence the biomechanical response of soccer players and thus may be used with caution to prevent injuries from occurring, particularly during preseason. This should help lower the overall incidence of injury experienced compared to other sports and help reduce the physical, financial, emotional and psychological effects of injury on the performer.

Appendix A: Example of the Prosole cushioning insole claims





Appendix B: Footscan Pressure Insoles: Accuracy and Reliability of force and pressure measurements

Abstract

Pressure insole technology is commonly used in both clinical and research situations. The application of the pressure insole data is influenced by the accuracy and reliability of the insole data. To measure the accuracy and reliability of the Footscan pressure insole (500 Hz Rsscan, Belgium), eight participants (25.3 ± 4.3 yrs; 79.6 ± 8.5 kg) were used to test two pairs of pressure insoles. Four female participant tested pair one and four male tested pair two during eight running trials (3.8 m/s). Intraclass Correlation Coefficients (ICC) revealed that the reliability of the force and pressure data was excellent (ICC >0.7). The accuracy of the peak impact and propulsive force measurements taken with the insole was assessed by comparing the difference in magnitudes between the mean of eight trials collected via the insole and eight collected simultaneously with a force plate (AMTI, 500 Hz). These measurements were significantly lower with the insole than those measured on the force plate ($p < 0.01$). Therefore, despite the excellent reliability of measurements taken with the Footscan pressure insole, the accuracy of the impact and propulsive forces is low and any data collected should be used with caution, particularly if the aim is to use the data for a clinical purpose.

Introduction

The foot is one of the most ergonomically efficient structures of the body and can sustain enormous magnitudes of both forces and pressure [1]. However, there is convincing evidence to suggest that abnormal magnitudes and frequencies of both vertical impact forces [2, 3] and plantar pressures [1, 4, 5, 6, 7] are important biomechanical factors in the aetiology of lower limb injury and pathology.

Traditionally, the forces that occur during locomotion are measured with a force plate, and the impact and propulsive phase are commonly identified from the vertical ground reaction force data at heel strike and propulsion respectively. However, the magnitude of impact force is not a good estimate of the load travelling vertically through the heel as the peak force occurring during the impact phase is a combination of the heel force along with some mid-foot and fore-foot force [8]. Likewise, propulsive forces aim to detail forefoot loading, but does not detail the loading characteristics of specific structures of the forefoot where ailments such as plantar ulceration or stress fractures of the metatarsal can be experienced. Force plates are not able to detail this specific region load, nor can they measure the force at the most important location, that which occurs between the foot and the shoe. Force plates are also impractical to use under certain circumstances as they are often in a fixed in-door location and are difficult to move. They also only provide the forces for one step. Therefore, in order to measure multiple steps, multiple force plates are needed.

Advancements in microcomputer technology [9] have enabled manufacturers to successfully develop in-shoe devices that provide robust, versatile and flexible [10] measurements of plantar forces and pressures occurring at the shoe and foot interface [9, 11, 12] without compromising the normal foot movement [9]. Pressure insoles are constructed with many evenly distributed sensor cells. Each cell acts as an independent force measuring device, the sum of which is equivalent to the total force applied to the body [13] and should equal the vertical force vector if measured simultaneously with a force plate. The advantage of the pressure insole is that because of the cell measurement of force, the calculation of the plantar pressure can be made based on the measured vertical load and the cell area [14]. Pressure can therefore be calculated for one cell, but can also be calculated for larger areas by identifying a region of interest, calculating the sum of the force from the cells within the region and then dividing it by the area the sensors cover.

Pressure insoles have been utilised by clinicians and researchers alike, in fields such as sport and exercise science, orthopaedics, diabetes and rheumatology [14] for clinical screening, treatment, behavioural modification [10] and clinical diagnosis [15]. The success of their application is dependant upon the systems validity [9, 16]. For pressure insoles to be used, a realistic measurement of plantar forces and pressures should be provided [9] which are repeatable so that to ensure consistency and reliability over time [17].

Previous studies have quantified the accuracy and reliability of various pressure insole models such as the F-scan, Novel Pedar and Biofoot insoles [9, 12, 18]. The Footscan® pressure insole system (Rsscan, Belgium) has been used with running participants [19, 20, 21, 22, 23]. However, to the authors' knowledge and despite its use in research, no thorough investigation has been published regarding the accuracy and reliability of the

Footscan pressure insole system. In addition, previous reliability studies have reviewed one set of insoles and assumed that these results are applicable to all insoles made with that technology, which may not be the case. The aim of this investigation was therefore to calculate the accuracy and reliability of the Footscan pressure insole. The reliability and accuracy of the Footscan insole was assessed using two pairs of Footscan pressure insoles. The reliability was assessed when the data sets were combined from each insole as well as when looking at the data reliability of measurements taken from each pair of insoles and for each individual insole (left and right).

Method

Eight participants (4 female and 4 male) (25.3 ± 4.3 yrs; 79.6 ± 8.5 kg) were used to assess the reliability and accuracy of the Footscan pressure insole (RSscan International, Belgium, 500 Hz). Plantar forces and pressures were collected from two new pairs of Footscan pressure insoles, of different shoe size (pair 1, sizes 5-6; pair 2, sizes 9-10). The four female participants were used to test pair one and four male participants were used to test pair two. Each pair consisted of insoles that were thin (0.7mm) and flexible, with piezoelectric polymer sensors of 7x5mm size and a resolution of 4 sensors/cm^2 (Figure 1). The insoles were placed flat within a pair of plimsolls and were connected to a data logger worn around the waist of the participant.

Figure 1: **Example of a pressure insole worn by participants**

A memory card capable of collecting 7 seconds of data was inserted in to the data logger. After each trial, data were transferred to a laptop computer and analysed in the Footscan 2.39 software package.

Procedure

Participants were asked to perform 16 running trials at a speed of 3.81 m/s (± 5%). The running speed was monitored by two photosensitive timing gates positioned 1.5 metres either side a force plate which participants were asked to run across (AMTI, 500 Hz), to provide simultaneous force plate and insole measurements. The foot that made contact with the force plate was alternated for each trial (i.e. the first step was right, the second step was left etc) so that 8 right foot steps and 8 left foot steps were collected with the force plate and the pressure insole simultaneously. As the insole can collect multiple steps in one trial, the pressure insole data that was analysed corresponded to the step which contacted the force plate. The accuracy and reliability of the vertical force measurements can be influenced by the temperature [18, 24], so therefore temperature was taken from within the plimsoll whilst being worn via a mercury thermometer for 2 minutes at the beginning, middle and end of the data collection session.

Measured Parameters

For each analysed step, masks were set for 7 different regions. These were based according to the software's automatic placement of pressure masks at the medial (M) and lateral (L) heel and the five metatarsals (1-5). From these masks, peak pressures were recorded. Likewise, peak impact force and peak propulsive force were taken from the resultant force calculated with the insole and from the vertical ground reaction force calculated by the force plate.

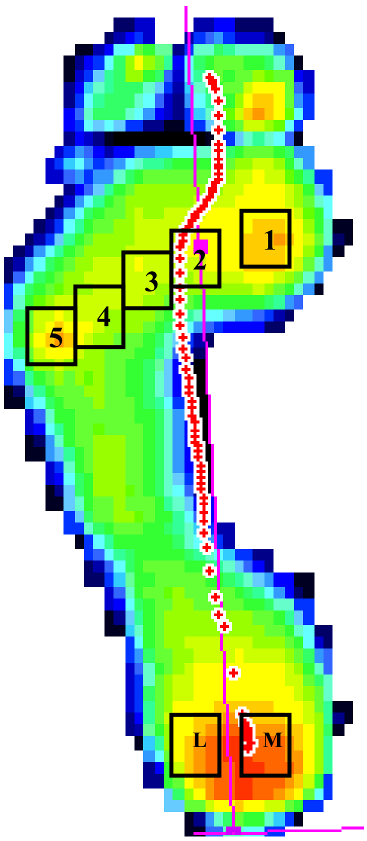


Figure 2: **Location of the pressure location**

Data reliability and accuracy

The reliability of the pressure insole data was determined by calculating an Intraclass Correlation Coefficient (ICC) on the data provided during eight trials for all participants. The size of the variation within the eight trials was also reported as a coefficient of variance (CoV) which reports the variance of the standard deviation from the mean as a percentage. The accuracy of the pressure insole to record impact and propulsive force was assessed by comparing the magnitude of impact and propulsive forces attained from the force plate to those measured by the insole. A paired samples t-test was performed between the means of eight trials taken with the insoles to the means of the eight trials measured by the force plate. The data was also divided and assess according to the size of the insole and whether it was the left or right insole (Figure 3). The alpha level was set at 0.05.

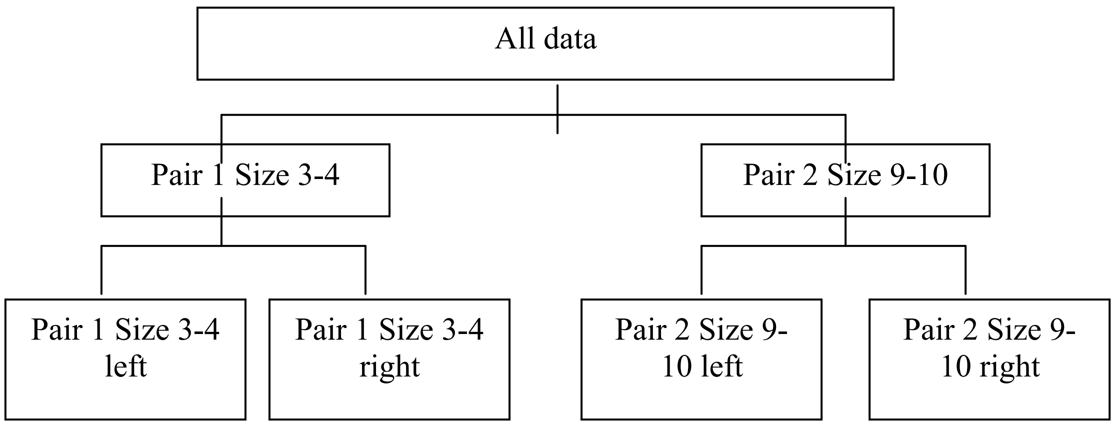


Figure 3: **An example of how the data was divided and analysed.**

Results

The temperature of the shoe was taken at the beginning, middle and the end of the collection of force and pressure data. For each participant, the temperature did not differ being a consistent 36.5°.

The size of the ICC coefficients indicated that excellent reliability was achieved for all measured parameters (ICC > 0.7) when the data from each insole and participant was combined. Good to excellent reliability (ICC 0.5-0.7) was also observed for much of the data when separated into the insole pairs and when the data for each pair was separated into the left and right insole. However, poor measurement reliability was observed for peak pressure measurements at metatarsal one and three (right insole of pair one) and peak impact force (left insole of pair two). Analysis of the reliability of peak pressure measurement at metatarsal two was also poor for the left insole of pair two as was the reliability of the peak pressure measurements at the second metatarsal when the data provided by both left and right insoles was combined (Table 1).

Table 1: *Summary of the Intra-class Correlation Coefficients (ICC)*

	All	**Pair 1**	**Pair 1 right**	**Pair 1 left**	**Pair 2**	**Pair 2 right**	**Pair 2 left**
Peak impact force	0.89*	0.91*	0.92*	0.93*	0.51*	0.68*	-0.07
Peak propulsive force	0.75*	0.75*	0.91*	0.93*	0.62*	0.57*	0.57*
Peak MTP1 pressure	0.75*	0.78*	0.38*	0.74*	0.67*	0.74*	0.71*
Peak M2 pressure	0.93*	0.91*	0.94*	0.90*	0.37*	0.51*	0.12*
Peak M3 pressure	0.81*	0.81*	0.11*	0.83*	0.65*	0.53*	0.84*
Peak M4 pressure	0.82*	0.82*	0.72*	0.91*	0.74*	0.42*	0.92*
Peak M5 pressure	0.81*	0.71*	0.68*	0.78*	0.91*	0.92*	0.95*
Peak H1 pressure	0.87*	0.72*	0.92*	0.86*	0.88*	0.79*	0.58*
Peak H2 pressure	0.80*	0.66*	0.92*	0.83*	0.85*	0.91*	0.61*

* = $p<0.05$

Table 2 provides a summary of the coefficients of variance, indicating the percentage variance between the magnitudes of the eight measurements. This showed that the CoV values ranged between 4.67 for the propulsive force (pair two), and 28.29 percent for the measurement of peak pressure at the lateral heel (left insole from pair one).

Table 2: *Summary of the Coefficient of variance (%) over eight trials*

	all	**Pair 1**	**Pair 1 right**	**Pair 1 left**	**Pair 2**	**Pair 2 right**	**Pair 2 left**
Peak impact force	11.07	8.64	9.73	8.85	12.76	15.93	8.72
Peak propulsive force	7.78	12.30	13.54	11.10	4.67	5.93	3.05
Peak MTP1 pressure	14.60	11.84	16.52	17.03	16.84	12.33	11.11
Peak M2 pressure	10.05	9.23	11.38	9.61	10.46	11.86	6.00
Peak M3 pressure	11.83	11.19	14.05	10.59	12.29	14.48	7.36
Peak M4 pressure	10.85	13.01	14.82	11.05	7.96	9.22	6.36
Peak M5 pressure	13.45	15.82	14.97	16.72	9.76	10.10	9.33
Peak H1 pressure	17.36	18.41	15.15	18.32	16.75	21.58	14.78
Peak H2 pressure	20.94	16.40	24.30	28.29	26.02	17.73	15.24

Table 3 shows the difference between the insole and force plate measurement for the peak impact and propulsive forces. The mean of the peak impact and propulsive force measurements for the eight left and eight right steps collected with the insole were significantly lower than those same steps measured by the AMTI force plate. The percentage difference ranged from 38.42 to 57.3 % depending on the measurement. Typically the difference in both impact and propulsive forces were less with pair one than with pair two.

Table 3: *Mean impact and propulsive forces attained with the insoles and force plate, the percentage difference and the significance of the paired samples t-test*

		Insole	**Force plate**	**Percentage difference %**	**P**
Peak Impact Force (N)	**All**	686.96 (± 173.82)	1372.30 (± 394.46)	49.94	0.01*
	Pair 1	618.96 (± 57.44)	1030.54 (± 121.70)	39.94	0.01*
	Left	609.25 (± 59.27)	1040.25 (± 148.79)	41.44	0.01*
	Right	628.66 (± 55.62)	1020.83 (±110.31)	38.42	0.01*
	Pair 2	755.31 (± 96.01)	1714.06 (± 227.25)	55.93	0.01*
	Left	737.41 (± 117.45)	1617.2 (± 147.51)	54.4	0.01*
	Right	773.20 (± 67.43)	1810.92 (± 271.52)	57.3	0.01*
Peak Propulsive Force (N)	**All**	856.39(± 171.42)	1664.88 (± 377.36)	48.56	0.01*
	Pair 1	760.14 (± 93.53)	1355.36 (± 59.97)	43.92	0.01*
	Left	750.33 (± 101.62)	1364.82 (± 83.66)	45.02	0.01*
	Right	769.95 (± 85.43)	1345.89 (± 33.96)	42.79	0.01*
	Pair 2	952.64 (± 44.39)	1974.40 (± 287.34)	51.75	0.01*
	Left	936.63 (± 55.54)	1976.25 (± 343.79)	52.61	0.01*
	Right	968.65 (± 29.53)	1972.55 (± 272.85)	50.89	0.01*

* = p<0.01

Discussion

The reliability and accuracy of a force and pressure measuring device is crucial when the data provided by the instrument is used for clinical and research related assessment. Previously, the accuracy and reliability analysis has been made for different pressure measuring devices such as the Pedar, Biofoot and F-Scan insoles. However, to the author's knowledge, prior to this paper no published research has been performed regarding the reliability and accuracy of the Footscan pressure insole, despite its use in clinical and biomechanical studies.

To assess the reliability of the Footscan pressure insole, an ICC was calculated for 8 measurements of peak impact and propulsive forces and peak pressures at the medial and lateral heel and the first to fifth metatarsals. This data was collected from the left and right feet of 8 participants. The reliability was assessed on the combined data from the left and right feet from two different pairs of insoles. It was thought that the quality of the insole data may be reduced as the insole ages as the sensitivity may be lost due to sensor damage. Therefore, by investigating the reliability of the insole technology in this way, insole damage is limited. The data was also divided and the individual insole reliability was further calculated. This quantified the reliability of the insole data from different insoles. This is particularly useful when the results from this study are to be applied to other pairs of Footscan insole with the assumption that similar reliability would be observed.

In comparison to published standards [9, 12], the size of the ICC coefficients provided from the combined data indicated that excellent reliability was achieved for all measured parameters. Good to excellent reliability was also observed for both pairs of insoles and for both the left and right insole in each pair. The only exceptions were that poor reliability was observed for the measurement of peak pressure at metatarsal one and three (the right insole of pair one), the measurement of peak impact force (left insole of pair two) and the measurement of peak pressures at metatarsal two (left insole of pair two; pair two). The current findings are also similar to those found with other insoles brand. Results using the F-scan®, Pedar® and BioFoot® have all shown that the reliability of peak pressures at metatarsals region is often good to excellent [9, 12, 25] and is commonly the most reliable of all the regions. Similarly, good to excellent peak pressure reliability have also experienced at the heel [12, 25] and for the reliability of the peak impact and propulsive forces [25].

The high reliability attained by the Footscan pressure insole suggests that the data produced is consistent under the testing procedures of this investigation. However, unlike the use of force plate, pressure insoles are also affected by other factors. The ability of the participant to consistently replicate the movement may influence the reliability of the pressure data measurements. Differences in insole humidity and temperature, as well as the insoles ability to maintain a high level of sensitivity over greater number of trials, with a more lengthily loading time of the sensors (> 1 hr), and day to day testing, have also all shown to effect pressure insole reliability [9, 26]. These factors may have influenced the reliability of the data, and explain the occasional poor reliability for some measurements. However, consistent in-shoe temperatures, the use of two pairs of insoles and short test duration may have reduced the effect of these factors in the current investigation. However, in circumstances where these factors can not be avoided, the effect of these factors must be understood and therefore some future research consideration is required. The high reliability of most insole measurements independent of the insole used to collect the data suggests that the reliability of insole measurements can also be applied to other Footscan pressure insoles. However, as shown by the poor reliability with smaller sample size, the data does highlight that occasionally the quality of the data can be poor reducing the reliability of some insole data. Therefore some caution is needed when making research conclusion, particularly if small sample sizes are used.

Despite knowing the size of the correlation via the measurement of the ICC the measurement does not indicate the variability within each calculated measurement. The CoV values taken indicated that the percentage variance between the magnitudes of the eight measurements ranged between 4.7% for the propulsive force (pair two), and 28.3% for the measurement of peak pressure at the lateral heel (left insole of pair one), which are values similar to those in previous research [7, 12]. In particular, the calculated CoV amongst the pressure measurements at the metatarsal have been reported at 15 and 15.5% [12] which is within the range found in the present study for all metatarsals accept the fifth. The CoV of the heel masks were greater than those presented for the metatarsals and the impact and propulsive forces. This may have been because the mask sizes were too small and as such small changes in heel placement may result in a relocation of the force at the heel away from the masks, and thus cause greater variability between the measurements. Increasing the mask size may decrease the variability between each measurement at the heel, although this may smooth any meaningful change in plantar pressure.

Another important characteristic of the pressure insole is the measurement accuracy. Despite the good reliability of the impact and propulsive forces measurements, these measurements were significantly less than those measured on the AMTI force plate. Hurkmans et al. [18], Boyd et al. [25], Barnett et al. [27] and Kalpen and Seitz [28] compared the vertical forces attained with a force plate with the forces experienced by the pressure insoles and found that in general, the forces collected by the insole system were consistently less than those of the force plate which supports the trends presented in the current investigation. However, the differences between the pressure insole and the force plate in the current investigation are greater than those reported by previously [25, 27, 28]. Kalpen and Seitz [27] reported the values collected with pressure insoles were 14-16% lower compared to force plate data whereas values of 38.42-57.3% were shown between the pressure insoles and the force plate measurements taken in the current investigation.

Like the reliability of the measurements, the temperature and humidity in the shoe and loading of the sensors during an entire day has been highlighted as influential when making accurate vertical force measurements with a pressure insole system [10, 18, 29] especially when the sensors are constructed from a piezoelectric crystal [29]. Since the current study showed that the temperature measured with a thermometer did not differ over time, and that the data collection was over a short period of time (>1hr) the loading of the sensors may not have influence the accuracy of the data and therefore this is not suspected to have caused the difference. Another explanation given by these authors was that the insoles sensors measure force "normal" to each sensor in the matrix, which is not necessarily the same as the vertical ground reaction force [18, 28, 30, 31]. During walking Barnett et al. [27] explained that the angle of the foot at heel strike or impact influences the angle of the force vector. As a result during the initial of the stance phase, the vectors are altered compared to the vertical force on the force plate. Kalpen and Seitz [28] showed that this was more obvious during the impact phase than during the propulsive phase and as a consequence the sensors do not measure all of the force occurring at impact and underestimates the forces that occur. On the other hand the measurements of the propulsive force are often more similar. Boyd et al. [25] published values of 6% and Barnett et al. [27] 3-11 %, but despite these findings, the difference in peak propulsive force measurement in the current investigation was greater than reported at impact. This observation along with the evidence that the differences between the insole and force plate for the measurement of both impact and propulsive forces was larger than found in previous investigations, may relate to the running task, which may have a greater effect on the ability to measure forces compared to walking. Further still, the loss of sensitivity may possibly relate to the position of the pressure sensors in neat rows across the insole. Between each sensor is an area of "dead space" [28] that will experience some of the load carried by the wearer, and which will also contribute to the underestimation of peak impact and propulsive forces. It was also observed that the difference in measurements was greater with in insole pair two than pair one. This may relate to size of the insole and that although more sensors are supplied in larger insoles, the proportion of force not being measured because of "dead space" and angle of the foot may increase, thus causing a larger difference between these insoles and the force plate. This speculation may be further supported by the fact that similar differences were observed for the left and right insoles within each pair. Therefore, the choice of insole size may ultimately influence the accuracy of any research conclusions or clinical diagnosis made

In conclusion, the Footscan pressure insole provides reliable force and plantar pressure data when eight running trials are collected. However, as pressure insoles can record multiple steps, more understanding of the reliability of insole measurements taken when eight consecutive steps are collected in one trial is necessary. This would enable both clinicians and researchers using this pressure insole to reduce data collection time. Likewise the ability for the insole to provide measurements that are repeatable on different days would ensure that the insole was consistent over time. However, it was observed that despite the reliability of the insole measurements the accuracy of the peak impact and propulsive forces taken with the insole was significantly less than measured by a force plate. Therefore, the use of the Footscan pressure insole system maybe more appropriate when used during comparisons of different conditions rather than when absolute magnitudes are needed for clinical purposes.

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Appendix C: Turning movement data reliability

Introduction

When collecting biomechanical data, it is important that the values obtained are representative of those that typically occur by the participant. As such, multiple trials are collected and the mean of these trials are used to compare independent variables. However, this mean is only representative if there is consistency between the performance of multiple trials. As such, knowledge of how many trials are required for representative data is required.

During running the reliability of force and pressure data collected was good to excellent for most measured parameter. However, Stacoff et al. (1988) found that the variation during lateral movements was double than that observed during running. Thus, it was deemed important to assess the consistency of data provided during turning movements.

Methods

Ten participants were used to collect data from eight right foot turning movements. This movement was performed by participants who ran up to a marked area which was 1m^2. They then placed their right foot into marked area, twisted their hips and pushed of in the direction in which they came. The speed of the turning movement was consistent for each turn, but was specific to the individual participant. This was monitored by a timing gate positioned one meter before the marked area. Timing started when an infra-red light beam was broken by the participant when approaching the plate, and stopped when the beam was broken for a second time when participants ran back in the direction in which they came.

Participants wore a pair of Footscan pressure insoles within a pair of moulded soccer boots (Adidas, Copa Mundial). The reliability of the data provided by the right pressure insole was assessed using an Intraclass Correlation Coefficient (ICC) in the eight trials. ICC measurements were taken for the selected parameters of peak impact force, peak propulsive force, peak pressure (medial and lateral heel and first and fifth metatarsal) and peak pressure loading rate (medial and lateral heel and first and fifth metatarsal).

Results and Discussion

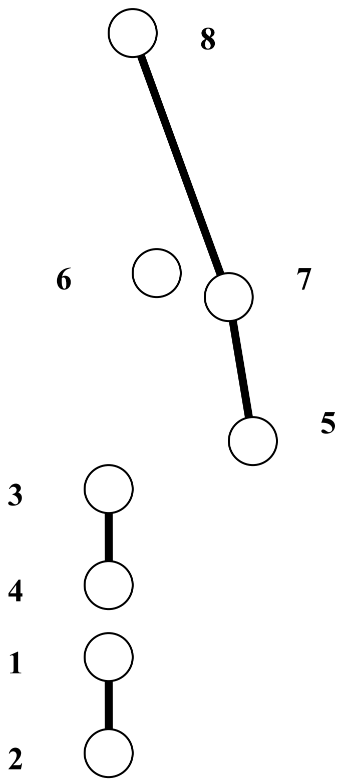
Table 1 *Results of the reliability tests on data collected with the pressure insole for peak impact force, peak propulsive force, peak pressure (medial and lateral heel and first and fifth metatarsal) and peak pressure loading rate (medial and lateral heel and first and fifth metatarsal).*

Variable	ICC	P
Peak Impact Force	0.88	0.001
Peak Propulsive Force	0.78	0.001
Peak Medial Heel Pressure	0.86	0.001
Peak Lateral Heel Pressure	0.90	0.001
Peak First Metatarsal Heel Pressure	0.87	0.001
Peak Fifth Metatarsal Heel Pressure	0.97	0.001
Peak Medial Heel Pressure loading rate	0.84	0.001
Peak Lateral Heel Pressure loading rate	0.90	0.001
Peak First Metatarsal Heel Pressure loading rate	0.62	0.001
Peak Fifth Metatarsal Heel Pressure loading rate	0.89	0.001

The results of the reliability analysis of the data provided by the footscan pressure insole during turning for each parameter showed that good to excellent reliability was obtained for all measurements (ICC > 0.07). This suggests that the data provided over 8 turning movements was reliable, thus the mean produced was reliable. As this is the same number of trials needed for good to excellent reliability for data collected with the Footscan pressure insole during running (Appendix B), similar number of trials are required for reliable data during running and turning movements used to compare different independent variables such as footwear and playing surfaces.

Appendix D: An example of the markers and segments used to calculate three dimensional joint angles

1. Calc1
2. Calc2
3. Ach1
4. Ach2
5. Shin
6. Medial knee
7. Lateral knee
8. Hip



Ankle dorsi-flexion/ plantar flexion: Degrees [ASIN[DOT[pron, z_shank]]]

Inversion/ eversion: Degrees [ASIN[DOT[x_shank, z_foot]]]

Knee flexion/ extension: Degrees [ASIN[DOT[knee abad, z_thigh]]] pron: Calculate unit vector as a cross product between x shank vector and z foot vector

x_shank: Calculate unit vector as a cross-product between z shank vector and ach2-ankle vector.

z_foot: Calculate unit vector directed from calc2 point to calc1 point.

z_shank: Calculate unit vector directed from Ach2 point to Ach1 point.

Ach2-ankle: Calculate unit vector directed from ach2 point to shin point.

knee_abad: Calculate unit vector as a cross product between Knee flexext vector and z shank vector

knee_flexext: Calculate unit vector directed from knee med point to knee lat point.

z_thigh: Calculate unit vector directed from knee lat point to hip point.

Appendix E: An example of the MatLab code written to calculation ankle joint moments and Achilles tendon forces

```
Clear

%input kinematic data file
fname=input('filename (kinematic):  ','s');
M = dlmread(fname)
%input force data file
j=find(fname=='.');
fname2=input('filename (force):  ','s');
N = dlmread(fname2)
k=find(fname2=='.');
[rforce, cforce] = size(N);
% inertia characteristics (foot weight - Wf; foot mass - mf; malleolus height - mallHt;
malleolus width - mallWt;
% foot length - Lf; forefoot width - ffWt; moments of inertia - Ix, Iy, Iz)
BW = input('subject bodyweight (N): ')
mallHt = 0.07;
mallWt = 0.05;
Lf = 0.20;
ffWt = 0.12;
% foot mass (mf) shank mass (msh)
mf = (0.0083*BW/9.81) + (254.5*Lf*mallHt*mallWt);
msh = 0.043*BW/9.81;
Ix = (0.00023 * mf) * ((4*mallHt^2) + (3*Lf^2)) + 0.00022;
Iy = (0.0041 * mf) * (mallHt^2 + ffWt^2) - 0.00008;
Iz = (0.00021 * mf) * ((4*ffWt^2)+(3*Lf^2)) + 0.00067;
Ishx = 0.0504;
Ishy = 0.05;
Ishz = 0.0005;

Fx=N(:,1); Fy=N(:,2); Fz=N(:,3); ax=N(:,7); ay=N(:,8); Tz=N(:,9);
% interpolate force data from 960Hz to 120Hz
t = 1:1:rforce;
```

```
x = 0:8:(rforce);
yFz = interp1(t, Fz, x, 'spline');
yFx = interp1(t, Fx, x, 'spline');
yFy = interp1(t, Fy, x, 'spline');
yax = interp1(t, ax, x, 'spline');
yay = interp1(t, ay, x, 'spline');
yTz = interp1(t, Tz, x, 'spline');
% size of new force vector
[cforce2, rforce2]= size(yFz);
% calculate heel strike and toe-off times
TO=0;
Time=0;
Cont = 0;
for i=1:rforce2;
    if yFz(i) > 20
   Cont=t(i);
    break
   end
end
for i=Cont:rforce2;
if yFz(i) < 20
   TO=t(i)
   break
   end
end
Total = (TO - Cont)/120;
for i=Cont:TO;
   T(i)= (1/120)*i;
end
% calculate baseline vertical force magnitude (zero)
baseline = mean(yFz(1:(Cont)-2));
```

```
P1=M(:,1:3); P2=M(:,5:7); P3=M(:,9:11); P4=M(:,13:15); P5=M(:,17:19);
P6=M(:,21:23); P7=M(:,25:27); P9=M(:,33:35); P10=M(:,37:39);
P11=M(:,41:43); P12=M(:,45:47); P13=M(:,49:51); P14=M(:,53:55); P16=M(:,61:63);
 % calculation of moment arm length
 for i=Cont:TO;
    d(i) = norm(cross(P3(i,:)- P4(i,:), P13(i,:)-P4(i,:)))/norm(P3(i,:)-P4(i,:))-0.007-
0.00429;
 end
% caculate foot CoM
for i =Cont:TO+2;
   comfx(i) = P2(i,1) + 0.449*(P14(i,1)-P2(i,1));
   comfy(i)= P2(i,2) + 0.449*(P14(i,2)-P2(i,2));
   comfz(i)=P2(i,3) + 0.449*(P14(i,3)-P2(i,3));
end
% calculate foot and shank unit vectors
z_foot = (P2-P1);
z_foot_inv = (P2-P1)';
vect2_foot = (P9-P2);
vect2_foot_inv = (P9-P2)';
vect3_foot = (P9-P1);
vect3_foot_inv = (P9-P1)';

for i = Cont:TO+2;
  v(i) = z_foot(i,:)*z_foot_inv(:,i);
  z_foot_unit(i,:) = (z_foot(i,:))/(sqrt(v(i)));
  v(i) = vect2_foot(i,:)*vect2_foot_inv(:,i);
  vect2_foot_unit(i,:) = (vect2_foot(i,:))/(sqrt(v(i)));
  v(i) = vect3_foot(i,:)*vect3_foot_inv(:,i);
  vect3_foot_unit(i,:) = (vect3_foot(i,:))/(sqrt(v(i)));
end
x_foot_unit = cross(vect2_foot_unit,vect3_foot_unit);
y_foot_unit = cross(x_foot_unit,z_foot_unit);
z_shank = (P4-P3);
z_shank_inv = (P4-P3)';
vect2_shank = (P5-P3);
```

```
vect2_shank_inv = (P5-P3)';
vect3_shank = (P5-P4);
vect3_shank_inv = (P5-P4)';
for i = Cont:TO+2;
  v(i) = z_shank(i,:)*z_shank_inv(:,i);
  z_shank_unit(i,:) = (z_shank(i,:))/(sqrt(v(i)));
  v(i) = vect2_shank(i,:)*vect2_shank_inv(:,i);
  vect2_shank_unit(i,:) = (vect2_shank(i,:))/(sqrt(v(i)));
  v(i) = vect3_shank(i,:)*vect3_shank_inv(:,i);
  vect3_shank_unit(i,:) = (vect3_shank(i,:))/(sqrt(v(i)));
end
x_shank_unit = cross(vect2_shank_unit,vect3_shank_unit);
y_shank_unit = cross(x_shank_unit,z_shank_unit);

% calculate ankle joint angles
e1 = x_shank_unit; %flex-ext axis
e2 = cross(x_shank_unit,z_foot_unit); %inv-ev axis
e3 = z_foot_unit; %ab-ad axis

for i = Cont:TO+2;
  dot_productdf(i) = sum((z_shank_unit(i,:)).*(e2(i,:)));
  df_angle(i)=(180/pi)*(asin(dot_productdf(i)));
  dot_productinev(i) = sum((x_shank_unit(i,:)).*(e3(i,:)));
  in_ev_angle(i)=(180/pi)*(asin(dot_productinev(i)));
  dot_productabad(i) = sum((x_foot_unit(i,:)).*(e2(i,:)));
  ab_ad_angle(i)=(180/pi)*(asin(dot_productabad(i)));
end
% calculation of components of transformation matrix for foot segment (Cardanian
angles)
for i=Cont:TO+2
t31(i) = z_foot_unit(i,1);
t32(i) = z_foot_unit(i,2);
t33(i) = z_foot_unit(i,3);
t21(i) = y_foot_unit(i,1);
t22(i) = y_foot_unit(i,2);
```

```
t23(i) = y_foot_unit(i,3);
t11(i) = x_foot_unit(i,1);
t12(i) = x_foot_unit(i,2);
t13(i) = x_foot_unit(i,3);
theta(i) = asin(t31(i));
theta_deg(i) = (180/pi)*theta(i);
end

for i=Cont:TO+2;
  y(i) = -t32(i)/t33(i);
  thi(i) = atan(y(i));
  y(i) = -t21(i)/t11(i);
  psi(i) = atan(y(i));
end
%angular velocities for foot segment
theta_vel = (diff(theta))/120;
thi_vel = (diff(thi))/120;
psi_vel = (diff(psi))/120

for i = Cont:TO;
  wx(i) = ((theta_vel(i))*sin(psi(i))) + ((thi_vel(i))*cos(theta(i))*cos(psi(i)));
  wy(i) = ((theta_vel(i))*cos(psi(i))) - ((thi_vel(i))*cos(theta(i))*sin(psi(i)));
  wz(i) = ((thi_vel(i))*sin(theta(i))) + psi_vel(i);
end

%angular accelerations for foot segment
theta_acc = (diff(theta_vel))/120;
thi_acc= (diff(thi_vel))/120;
psi_acc= (diff(psi_vel))/120;

for i = Cont:TO;
  wwx(i) = (thi_acc(i)*cos(theta(i))*cos(psi(i))) -
(thi_vel(i)*theta_vel(i)*sin(theta(i))*cos(psi(i))) + (thi_vel(i)*psi_vel(i)*(-
sin(psi(i)))*(cos(theta(i)))) + (theta_acc(i)*sin(psi(i))) +
(theta_vel(i)*psi_vel(i)*cos(psi(i)));
```

```
  wwy(i) = (thi_acc(i)*cos(theta(i))*sin(psi(i))) +
(thi_vel(i)*theta_vel(i)*sin(theta(i))*sin(psi(i))) -
(thi_vel(i)*psi_vel(i)*(cos(psi(i)))*(cos(theta(i)))) + (theta_acc(i)*cos(psi(i))) +
(theta_vel(i)*psi_vel(i)*(-sin(psi(i))));
  wwz(i) = (thi_acc(i)*sin(theta(i))) + (thi_vel(i)*theta_vel(i)*cos(theta(i))) +
psi_acc(i);
end

% calculate ankle joint force
 % foot CoM acceleration components
 Xf = comfx'; Yf = comfy'; Zf = comfz';
 Xf_vel = (diff(Xf))/120; Yf_vel = (diff(Yf))/120; Zf_vel = (diff(Zf))/120;
 Xf_acc = (diff(Xf_vel))/120; Yf_acc = (diff(Yf_vel))/120; Zf_acc = (diff(Zf_vel))/120

 % ankle force components in global system
for i = Cont:TO;
  Fank_X(i) = (mf*Xf_acc(i)) - yFx(i)';
  Fank_Y(i) = (mf*Yf_acc(i)) - yFy(i)';
  Fank_Z(i) = (mf*Zf_acc(i)) - mf*9.81 - yFz(i)';
end
Fank = [Fank_X; Fank_Y; Fank_Z];
% ankle force components in local system
for i=Cont:TO;
T11(:,:,i) = cos(theta(i))*cos(psi(i));
T12(:,:,i) = sin(thi(i))*sin(theta(i))*cos(psi(i)) + (cos(thi(i))*sin(psi(i)));
T13(:,:,i) = -cos(thi(i))*sin(theta(i))*cos(psi(i)) +(sin(thi(i))*sin(psi(i)));
T21(:,:,i) = -cos(theta(i))*sin(psi(i));
T22(:,:,i) = -sin(thi(i))*sin(theta(i))*sin(psi(i)) + cos(thi(i))*cos(psi(i));
T23(:,:,i) = cos(thi(i))*sin(theta(i))*sin(psi(i)) + sin(thi(i))*cos(psi(i));
T31(:,:,i) = sin(theta(i));
T32(:,:,i) = sin(thi(i))*cos(theta(i));
T33(:,:,i) = cos(thi(i))*cos(theta(i));
end
for i=Cont:TO;
%T_global_local(:,:,i) = [t11(i) t12(i) t13(i); t21(i) t22(i) t23(i); t31(i) t32(i) t33(i)];
```

```
T_global_local(:,:,i) = [T11(:,:,i) T12(:,:,i) T21(:,:,i); T21(:,:,i) T22(:,:,i) T23(:,:,i);
T31(:,:,i) T32(:,:,i) T33(:,:,i)];
end
%Fank_local = (T_global_local)*(Fank);
Fank_x = Fank(1,:);
Fank_y = Fank(2,:);
Fank_z = Fank(3,:);
% foot moment arm calculations
for i = Cont:TO;
  dcopX(i) = yax(i) - P16(i,1);
  dcopY(i) = yay(i) - P16(i,2);
  dcopZ(i) = -P16(i,3);
  dankX(i) = P13(i,1) - P16(i,1);
  dankY(i) = P13(i,2) - P16(i,2);
  dankZ(i) = P13(i,3) - P16(i,3);
end
% foot weight
Wft = mf * 9.81;
% rate of change of linear momentum for foot CoM (Pft)
for i=1:TO;
Pft(:,:,i) = [mf*wwx(i); mf*wwy(i); mf*wwz(i)];
end
%rate of change of angular momentum
for i = Cont:TO;
W(:,:,i)=[0 -wz(i) wy(i); wz(i) 0 -wx(i); -wy(i) wx(i) 0];
dHdt(:,:,i)=[(Ix*wwx(i)); (Iy*wwy(i)); (Iz*wwz(i))];
w(:,:,i)=[wx(i);wy(i);wz(i)];
end
I=[Ix 0 0; 0 Iy 0; 0 0 Iz];
for i=Cont:TO;
H(:,:,i)=I*w(:,:,i);
Mcm(:,:,i)= (dHdt(:,:,i) + (W(:,:,i) * H(:,:,i)));
end
```

```
%define foot COM co-ords, ankle joint centre, ankle joint force

%Rcmfx=M(:,61);
%Rcmfy=M(:,62);
%Rcmfz=M(:,63);

Rcmfx=comfx;
Rcmfy=comfy;
Rcmfz=comfz;

Rdx=M(:,49);
Rdy=M(:,50);
Rdz=M(:,51);

%Fdt = transposed force matrix
Fdt = Fank.';
Fdx=Fdt(:,1);
Fdy=Fdt(:,2);
Fdz=Fdt(:,3);

% define matrices for COP(Rfp), foot COM(Rcmf), ankle joint centre(Rd), GRF(Ffp),
ankle joint
% force (Fd)

for i=Cont:TO;
Rfp(:,:,i) = [0 0 yay(i); 0 0 -yax(i); -yay(i) yax(i) 0];
end

for i=Cont:TO;
Rcmf(:,:,i)= [0 -Rcmfz(i) Rcmfy(i); Rcmfz(i) 0 -Rcmfx(i); -Rcmfy(i) Rcmfx(i) 0];
Rd(:,:,i)= [0 -Rdz(i) Rdy(i); Rdz(i) 0 -Rdx(i); -Rdy(i) Rdx(i) 0];
Ffp(:,:,i) = [yFy(i); yFy(i); yFz(i)];
end
```

```
for i=1:TO;
Fd(:,:,i)= [Fdx(i); Fdy(i); Fdz(i)];
end
% define free moment of GRF
for i=Cont:TO;
   RTz(:,:,i) = [0;0;Tz(i)];
end
%ANKLE MOMENT CALCULATION
for i=Cont:TO;
   Ma(:,:,i) = RTz(:,:,i) + ((Rfp(:,:,i) - Rcmf(:,:,i)) * Ffp(:,:,i)) -((Rcmf(:,:,i) - Rd(:,:,i)) *
Fd(:,:,i)) - Mcm(:,:,i);
end
for i=Cont:TO;
   Max(i)=Ma(1,1,i);
   May(i)=Ma(2,1,i);
   Maz(i)=Ma(3,1,i);
end
% transform from global to local co-ord frame
for i=Cont:TO;
   Ma_local(:,:,i) = (T_global_local(:,:,i))* (Ma(:,:,i));
end
for i=Cont:TO;
   Ma_localx(i)=Ma_local(1,1,i);
   Ma_localy(i)=Ma_local(2,1,i);
   Ma_localz(i)=Ma_local(3,1,i);
end
% identify time of peak moment values Ma_localx, Ma_localy, Ma_localz
PKMax = max(Ma_localx);
x = find(Ma_localx == PKMax);
TMPKMax = ( t(x) - Cont)/120;
PKMay = max(Ma_localy);
x = find(Ma_localy == PKMay);
TMPKMay = ( t(x) - Cont)/120;
```

```
PKMaz = min(Ma_localz);
x = find(Ma_localz == PKMaz);
TMPKMaz = ( t(x) - Cont)/120;
for i=Cont:TO
ATF(i) = Ma_localx(i)/d(i);
end
```

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