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A BIOMECHANICAL ANALYSIS OF NON-LINEAR MOTION IN SOCCER.

By Neal Antony Smith

Doctor of Philosophy

FACULTY OF SCIENCES

August 2000

This thesis has been completed as a requirement for a degree
of the University of Southampton.

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ABSTRACT

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This thesis has been completed as a
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Soccer consists of many different types of sports specific movement. The present level of understanding of non-linear motion is negligible yet required if improvements are to be made in technique and performance of such actions. This thesis aimed to establish mechanisms for non-linear motion relevant to soccer performance.

Preliminary analysis of curvilinear motion involved electromyographical analysis in selected muscles of the lower extremity at different grades of curvature. Results revealed adaptation of temporal muscle activity at the tightest grade of curvature. Adaptation occurred in both legs, but predominantly the outside leg, with increased duration of activity after footstrike (Smith et al., 1997). Stride kinematics were also altered, as increasing curve severity gave reduced stride length and increased stride frequency. Foot contact time was not changed as a function of curvilinear motion ($P > 0.05$), giving an increased proportion of the stride cycle in the stance phase. Rearfoot contact time increased as a function of curve severity ($P \leq 0.05$).

To describe and quantify adaptation of lower limb movement in curvilinear motion, three-dimensional kinematics were used. Subjects ($n = 8$) wore soccer footwear on natural turf. Ranges of motion at the lower extremity were increased at the faster of the two velocities tested (4.4 and 5.4 $ms^{-1} \pm 5\%$), yet tended to reduce with curve severity. The inside leg displayed more differences in angular displacement with curve severity, and the ankle joint showed to be a key adaptive site.

Ground reaction forces of two consecutive footfalls were performed on natural turf to assess relative contributions of the two limbs during straight and curvilinear motion at a 5m radius. Total force over two footfalls was greater during straight motion. A mechanism of lowered centre of gravity during curvilinear motion was proposed. During curvilinear motion the outside leg was associated with greater force values in all three planes, displaying a greater contribution to curvilinear motion.

Force measurements on natural turf were used to assess different sole configuration during three soccer specific moves. A modern moulded sole was found to be associated with greater maximum friction, also lower vertical ground reaction forces during a Cruyff turn and lower overall forces during the shot.

This thesis established biomechanical adaptations and suggested mechanisms during non-linear motion in the soccer player. The research represented the first experimental investigations in this area and therefore recommendations for future study are considered.

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ACKNOWLEDGEMENTS

As this thesis developed so many people were involved. This, therefore, makes thanking all of them individually very difficult. However, there are certain people who deserve special thanks for their assistance.

Firstly my subjects, without whose time and patience this study could not have been completed. Many of who have repeatedly given up their spare time to help make this work possible. In addition, the technical support I have received has made the experimentation process possible, and the majority of data collected would not have occurred without this assistance.

To my friends. However great or small a part you may think you have played in the completion of this work, I could not have done it without you, and I thank you for being there.

A special thanks goes to my family. Barry, Janet and Kay for their continuous support and encouragement, which has never wavered throughout my period of study. I just hope that the completion of this thesis goes someway towards showing a small part of the gratitude I owe to you.

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Thank you all.

CHAPTER 1

CHAPTER 1

Introduction

Soccer has been referred to as the world game, with over 200 million registered players (Federation Internationale Football Association, 1992). To many it is not only a game, but a way of life. Vast sums of commercial investment accompany today's professional soccer industry, leading to the development of a soccer culture within our society. As a result of increased participation and commercial interest, scientific investigation into soccer has steadily gained momentum to a stage where the sport has been regarded as a recognised avenue of scientific enquiry.

As an exercise modality, soccer is classified as that of high intensity, intermittent exercise superimposed on a background of physical endurance (Ekblom, 1986). The composition of the intermittent exercise covers of a range of discrete activities such as walking, jogging, and sprinting (forwards, backwards, sideways, diagonally), jumping, stopping, cutting, turning, pivoting and tackling. In addition to these physical demands within the game there are many skills to be mastered with the soccer ball, yet the summation of these skills accounts for only 1.73 % of the total distance covered by outfield players (Reilly and Thomas, 1976). The majority of the total distance covered during a game is achieved without the ball, helping to gain tactically or positionally beneficial positions on the field. Although time and motion studies (e.g. Reilly and Thomas, 1976; Yamanaka et al., 1988) of elite competition have examined the frequency of these activities, biomechanical analysis of non-linear locomotion is absent from the scientific literature on soccer. Although the breakdown of time performing straight and non-linear motion is not specified in the literature, the nature of the game would seem to demand some proportion of time to be given to performing non-linear actions. Such actions can range from curvilinear running in an overlapping move and during the approach to a free kick, to sharp direction changes to intercept a pass and dribble around one or more opponents. If the mechanisms for progression in these actions were known, the design of training to improve performance could be achieved. In addition, unique and valuable contributions to knowledge of the biomechanical demands of such actions could be gained. Thus, the aim of this thesis is to establish the mechanisms of non-linear motion specific to soccer performance.

Previous analysis of locomotion has taken the form of analysis of walking or running in a linear path overground, or on a treadmill. Investigations have focused on descriptive kinematics, particularly differences in technique at a range of speeds, and have been well documented by Cavanagh (1990). Additional information concerning muscular activity during linear motion has been gained through electromyography (Elliot and Blanksby, 1979; Schwab et al., 1983; MacIntyre and Robertson, 1987), detailing both temporal data and magnitudes of muscular activity at different speeds of locomotion. Through electromyographic analysis, the major muscles of the lower extremity involved in straight locomotion have also been identified (Elliot and Blanksby, 1979). The loading of the skeletal system during locomotion has also received considerable attention. The majority of studies have used force platforms to ascertain ground reaction forces acting on athletes at differing speeds (e.g. Cavanagh and Lafortune, 1980; Mann and Hagy, 1980; Clarke et al., 1983; Hamill et al., 1983; Munro et al., 1987), but all of these investigations were also concerned with linear motion. With the performance of non-linear motion so inherent to the game of soccer, the biomechanical analysis of non-linear motion was required to enable further understanding of the mechanisms involved in these movements.

The movements encountered during a game are typically situation dependent, with cutting, turning and pivoting occurring at differing speeds and angles, making standardisation of these movements difficult and limiting the ability of the biomechanist to understand the mechanisms that contribute to the successful performance of such tasks. Intuitively, it seems likely that underpinning effective performance of non-linear motion is the ability to sustain movement in a curvilinear path. Early pilot investigations by the author of this thesis revealed the frequency and severity of curvilinear motion in the professional game. For investigative purposes, curvilinear motion also represents a reproducible activity with which soccer players are familiar.

A theoretical analysis of motion along a curved path suggests this motion is dynamically distinctive from motion along a straight path (Dyson, 1968; Hay, 1978).

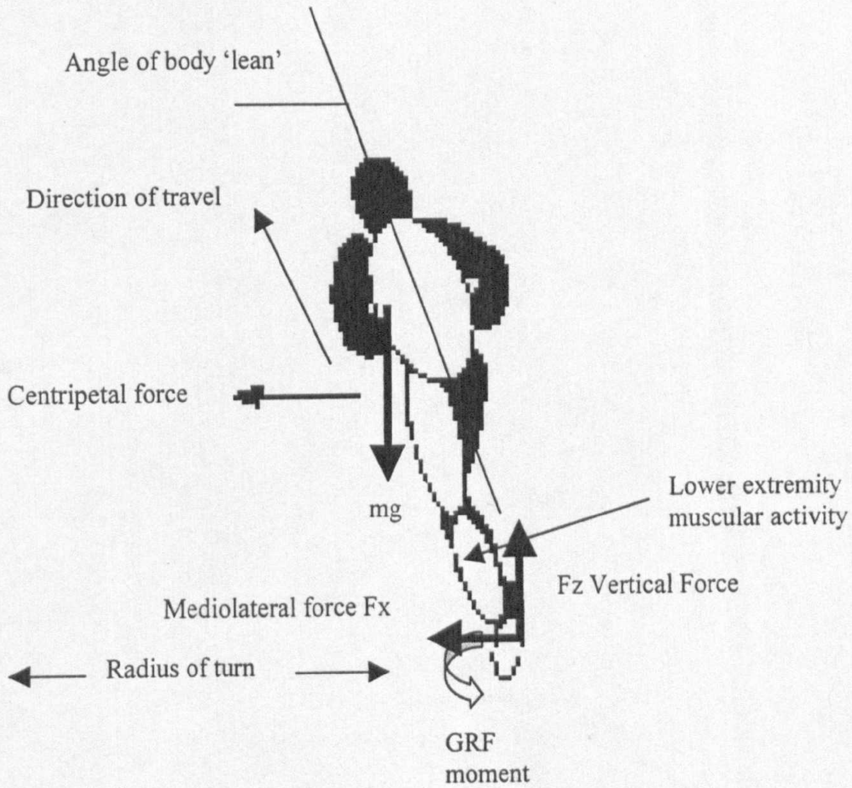

Figure 1.1 Diagram detailing factors affecting curvilinear motion (shaded leg represents the outside leg of the curve).

The mechanics of moving along a curved path indicates that runners must change their body positions, and thus adjust their lower extremity function, as they accomplish movements such as a track turn. Factors that will affect the athlete's movement are detailed in figure 1.1. The radius (or grade) of curvature required has an effect on the other variables described. Physical principles concerning movement in a circular path require generation of centripetal force. This force must be directed toward the centre of the circular path and can be calculated from the following equation;

$$C_f = \frac{mv^2}{r}$$

where the radius of the curved path is represented by (r). From figure 1.1 the force arrow Fx represents a mediolateral ground reaction force which will provide centripetal force. Such a mediolateral force could occur either at the inside or outside leg of the curve, or serially in both. If a mediolateral force were applied, a turning moment would be generated with respect to the body's centre of gravity. In figure 1.1, this would give the body a tendency to rotate about the centre of gravity in the frontal plane, in a clockwise direction. In fact, the body would not rotate about the centre of gravity, as anticlockwise body 'lean' counteracts the rotating moment. It can be deduced therefore that the different ground reaction force pattern must therefore be generated by altered muscular function within the lower extremity. This differing muscular activity at the lower extremity was thought by the author of this thesis to create the alternative path of movement seen in curvilinear motion. The use of telemetred electromyography signals would provide a method for quantifying the changes in applied muscular activity from a free roaming subject. For ecologically valid data relevant to soccer to be obtained, movement would need to be studied on a natural turf surface. This requirement creates additional problems for the valid and reliable measurement of ground reaction forces which will need to be resolved if useful information is to be obtained.

Although the muscles used in linear motion have been previously identified and monitored (Elliot and Blanksby, 1979; Schwab et al., 1983; MacIntyre and Robertson, 1987), one would expect additional muscles to be involved in non-linear motion, but these have not been detailed in the literature. To evaluate how the muscles of the lower extremity activate to enable curvilinear performance would require identification of those muscles through extensive pilot work. In addition, due to the differing displacement between the inside and outside legs during a curve, it would appear plausible that differences might exist within the temporal activity pattern of the specified muscles. Also, Andrews et al. (1977, cited by Schot et al., 1995) speculated that direction changes were accomplished in the main through torque generated by the torso, pelvis and lower extremity musculature being applied to the ground. Such postulations provide research questions that may be tested by an electromyographic experiment into curvilinear motion. Recent developments in biomechanical measurement techniques have led to the use of in-shoe pressure measurement devices; however, the use of such devices provides data on the effect of a particular movement, but does not provide information relating to the mechanisms causing the movement. The method of surface electromyography is a useful tool as it can provide information concerning the muscular activity leading to a movement. Yet, the muscular activity also reflects, and is moderated by, the perceptual feedback component generated by ground contact, and so might give a more valuable insight to the mechanisms taking place during curvilinear performance. The importance of perceptual feedback has been highlighted by studies into barefoot and shod running. Stockton and Dyson (1998) showed increased duration of muscular activity when running barefoot compared to shod. Such findings have shown that muscular activity differences with changing shoe-surface interface combinations can be detected by electromyography. Therefore, it was thought likely that the technique could be used to gain an insight into the muscular activity difference when soccer footwear is worn on a natural turf surface. Surface electromyography was therefore selected as a method for preliminary investigation into ergonomic ways of assessing non-linear motion.

De Wit et al. (1995) and Nigg (1990) showed that different kinematic measures could occur as a result of altered shoe-surface characteristics in running. In addition, Stuke et al. (1984) presented data displaying how an athlete responds kinematically to a difference in shoe-surface interaction during stopping. To verify such suggestion with respect to soccer, kinematic measures need to be taken. Basic stride length and stride frequency are kinematic parameters of locomotive performance which have received considerable attention in the running literature to date (eg. Dillman, 1975; Williams, 1985; Cavanagh, 1990). These, in addition to other kinematic measures of individual running technique have allowed researchers to understand the relative contributions of kinematic parameters in linear motion at different velocities. Any adaptations which may occur in these variables during curvilinear running, in addition to altered footwear conditions, could also help our understanding of the mechanisms involved in general non-linear motion in soccer.

From consideration of the Newtonian principles of motion, the reaction force from the ground can be important in our understanding of the body movement in a curvilinear path (see figure 1.1). However, soccer takes place predominantly on natural turf surfaces, especially at the top level (Winterbottom, 1985) which makes the measurement of ground reaction forces in an ecologically valid environment problematic. The need to measure ground reaction force under a standard ergonomical soccer interface of natural turf and studded footwear was identified during this research, and a suitable method was devised and validated. Consideration of these force measurements would also increase information available concerning the shoe-surface interface relationship.

The chronological importance and availability of scientific literature may affect advances in research and development. Progression of the research detailed in this thesis followed the rationale outlined above with respect to the collection of electromyography data. In addition, the specific effect of soccer footwear during curvilinear motion was investigated. Whilst a force platform rig which would allow the measurement of ground reaction force at the soccer boot-turf interface was under construction, a primary research paper in the little researched field of curvilinear motion was uncovered in the scientific literature.

In their paper which reported work on track running around a curve, Hamill et al. (1987) stated that curved path locomotion may subject individuals to unique stresses, and that research examining this type of movement had been largely neglected. They argued that progression in a path of constant curvature requires a series of cutting movements to occur. Such movements, it was suggested, achieve the change of direction by using one powerful step, and can occur using the outside leg (side-step), or the inside leg (cross-over step). A cyclical pattern of small cross-over cutting motions, followed by small side-step cutting motions enables the athlete to maintain a curvilinear path. They also reported ground reaction force variables at the inside and outside leg of a track turn. Their results, in addition to those of Stoner and Ben-Siri (1979) who studied sagittal plane kinematics of a runner on the curved portion of a track, raised research questions regarding the differing contribution of each limb to curvilinear progression, and also the quantification of the asymmetrical nature of curvilinear motion. The research outlined in this thesis further investigated the kinetics and kinematics during curvilinear motion on natural turf through ground reaction force measurements and three dimensional kinematics.

Due to the paucity of research in the area of non-linear motion, and the associated lack of recognised methods for its assessment, ergonomic methods of assessing non-linear motion with respect to soccer were investigated and developed in this thesis. Inherent in the ergonomic analysis of non-linear motion is the transfer of human kinetics at the shoe-surface interface. Ecologically valid experimentation relevant to soccer must also take account of any sports specific equipment in current use, and in soccer, the main equipment used during sports specific actions are the soccer boots. Whilst certain investigations have used soccer style footwear whilst measuring sagittal plane kinematics during curvilinear motion (Greene and McMahon, 1979), no studies exist which have examined the effect this footwear has on curvilinear performance. The inclusion of studded sections into the outsole of the footwear is intended to reduce slippage by increasing friction, and therefore effective force transmission to the ground. The effect of wearing such footwear on muscular activity and the ensuing ground reaction forces is important, but as yet remains unreserached. Soccer is characterised by sprinting, stopping, cutting and pivoting situations, where shoe-surface relations are critical, and frictional resistance must be within an optimal range. Soccer footwear predominantly uses six-studded outsoles, modern designs have incorporated angled 'blades' with an intention to replace traditional 'studded' designs, attempting to create greater translational friction during soccer movements. Soccer specific movements of turning and shooting present actions which place great stress on the shoe-surface interface, and can be used in an ecologically valid assessment of footwear outsoles. Research into quantifying the effects of altered shoe-surface interface in such non-linear motions is an area where greater knowledge may influence performance by providing unique information regarding the forces produced during these actions, and forms the research question of the final experiment in this thesis. This thesis attempts to investigate and improve knowledge concerning the constituent components that enable curvilinear motion, with the emphasis on actions which compose non-linear motion in soccer. The aim was to establish the mechanisms of non-linear motion specific to soccer performance. In addition, investigation was to provide information regarding how these actions are affected by specific shoe-surface interface characteristics.

Overview of Thesis

This thesis consists of eight chapters. In the first chapter the limited existing knowledge relating to the analysis of non-linear motion is highlighted, and the need for biomechanical analysis of soccer-related non-linear motion and sport specific movements considered. Chapter 2 provides a review of literature, drawing on biomechanical research conducted into soccer; related running literature, and scientific studies relating to the biomechanical techniques to be employed. The aim of this chapter is to outline relevant baseline data existing for linear motion, and to summarise work published on the biomechanics of soccer. The small amount of information available concerning non-linear, and more specifically curvilinear motion is also discussed. The chapter identifies where the intended research studies which form part of this thesis would improve existing knowledge. The weaknesses in the scientific methods reported in the literature are highlighted, along with the identification of areas where knowledge and experimental data are absent from the existing literature with respect to the questions raised in the introduction.

Chapter 3 reported on investigation of the muscular activity recorded in the lower extremity during running. Superficial muscles which were considered likely to be particularly involved in the production of curvilinear motion were identified and activity levels during linear and different grades of curvilinear motion were compared. Such data aimed to provide information regarding the muscular mechanisms of curvilinear progression when compared to straight motion. An understanding of the ergonomic conditions encountered at the shoe-surface interface in soccer, with the effect of sports specific footwear on curvilinear motion was required. Therefore, the perceptual-motor feedback adjustments likely to arise from different shoe-surface frictional properties were considered by comparing electromyographic activity when wearing soccer boots with that occurring wearing training shoes. Chapter 4 further investigated the changes in stride kinematics, as an increased amount of muscular activity around stance was identified in curvilinear motion. The experiment aimed to clarify the differences in movement of the inside and outside leg during curvilinear motion to aid explanation of the increased muscular activity around stance shown in Chapter 3. The adaptations to curvilinear motion in muscular activity reported in chapter 3 suggested that differing application of muscular force may cause altered accelerations in segments of the lower extremity and be responsible for the asymmetry exhibited in curvilinear motion. These muscular adaptations, in addition to the altered body position required to move in a curved path suggested altered kinematics during curvilinear motion. Chapter 5 aimed to quantify these adaptations to curvilinear motion in soccer footwear. The altered movement pattern during curvilinear motion also aimed to provide insight into the relative asymmetry of the inside and outside legs during curvilinear motion.

Asymmetry reported in curvilinear motion kinematics suggested differing functions of the two limbs. The segmental accelerations that occur to maintain curvilinear motion are transmitted through the footwear to the surface. Measurement of the ground reaction forces during consecutive footstrikes would enable quantification of the kinetic transfer occurring at the shoe-surface interface during curvilinear motion. Chapter 6 reports the results of ground reaction force measurement using a natural turf force plate rig development. Data concerning total force, mediolateral force and stance times aimed to provide further insight into the mechanisms of curvilinear motion and the relative contribution of each limb to curvilinear progression. The contrasting kinetic demands of straight to curvilinear motion were also compared over a full stride cycle. Differences between the two movement patterns infer key adaptations which may be trained to improve curvilinear performance in soccer players.

Modern developments in soccer footwear may influence the characteristics of the shoe-surface interface during non-linear soccer specific actions. Chapter 7 reports on footwear comparisons of performance in an ecologically valid natural turf environment using ground reaction force variables. Such results show how modern outsole design can affect performance during soccer specific actions. Finally, chapter 8 provides a summary and synthesis of the findings of completed research, discussing their implications for soccer performance and to sports biomechanics, also highlighting areas for future research.

CHAPTER 2

CHAPTER 2

Literature Review

2.1

OVERVIEW

Literature pertaining to the scientific study of locomotion in soccer draws on many diverse areas including physiology, biochemistry, and biomechanics. Contemporary work into the biomechanics of locomotion has focused on linear running, with applications to movement in soccer rare. Time and motion studies have been undertaken, and are reviwed, yet no information on non-linear motion has been provided in the scientific literature. Pilot work monitoring curvilinear movement during professional soccer revealed the pervading presence of non-linear motion in the game, yet the application of biomechanical investigation into soccer has focussed primarily on the assessment of kicking techniques. The findings on soccer kicking are relevant due to the use of soccer specific movements of kicking and turning in the final experimental chapter of this thesis. These are only briefly reviewed as the in-depth evaluation of soccer kicking is beyond the scope of this thesis. The small amount of literature concerned with other soccer specific movement patterns is also reported.

To examine the phenomenon of non-linear motion in soccer, three methods of analytical technique were required. Principally the methods used have been electromyography, cinematography/videography and ground reaction force analysis. Findings from these distinct investigative measures are summarised later in this review, yet primarily only encompass linear running. The key findings from biomechanical research in linear motion which have implications for non-linear movement in soccer are also reported. Each analytical technique used has its limitations supported by the literature, which are also reviewed along with any corresponding considerations for minimisation of these limitations.

An important component of non-linear motion in soccer is the interface between the athlete and the ground. In soccer, additional problems are posed due to the nature of the natural turf surface, and the specific footwear used. This review includes both the footwear and surface considerations of the interface which are relevant to experimental chapters. The continuing development of soccer footwear can have an impact on both extrinsic injury prevalence and also performance of soccer specific movements, therefore the effect of differing soccer footwear characteristics on gait performance is addressed also. Throughout the literature review, the limitations of the existing experimental studies reported in relation to the movements required during soccer play are considered.

2.2

SOCCER - HISTORICAL PERSPECTIVE AND SCIENTIFIC INVOLVEMENT

The modern sport of football has its origins in the middle of the nineteenth century when first records suggest it was played in English public schools. The first official rules for association football were published in 1848, with the Football Association the first to be founded in 1863 (Inklaar, 1994a). The Federation International of Football Associations (FIFA), founded 1904, involves 186 countries with a total of around 200 million licensed players (FIFA, 1992). The European soccer organisation (UEFA) represents 49 countries with 20 million licensed players (UEFA, 1988).

Outdoor soccer is played by both sexes. Competition occurs at adult level and also at youth level, where players compete in different age group categories. It is a team sport normally involving 10 field players and 1 goalkeeper in two opposing teams. The playing field has maximal dimensions of 68m x 105m with a surface generally of grass in higher level competition, but sometimes sand, gravel or artificial turf (Inklaar, 1994a). A regulation game will consist of two 45 minutes halves, separated by a 15 minute break at half time.

A soccer match provides an environment which places high stress upon individuals, with a wide variety of activities ranging from light exercise to maximal sprints. The average energy yield during a game of elite soccer is about 80% of the individual maximum. Physiologically, the exercise pattern can therefore be characterised as that of high intensity, intermittent non-continuous exercise (Ekblom, 1986) superimposed on a background of endurance exercise, with an average blood lactate concentration during a game of 7 to 8 mmol/l with peak values above 12 mmol/l. Most players have empty glycogen stores at the end of a game, are dehydrated and have increased body temperature (Ekblom, 1986). An English first division match was reported by Reilly and Thomas (1976) to contain approximately 900 discrete bouts of activity, or a change every 5 -6 seconds during the 90 minutes of play.

As soccer involves the amalgamation of several discrete activities such as walking, jogging, running, sprinting, jumping and turning, research itself has been multidisciplinary in nature. Scientific impetus gained from sprinting research as early as the fifth century B.C. (Cavanagh, 1990) has developed and combined with the development of distance running studies, and has been stimulated even further by the interest from large commercial equipment manufacturers. In more recent years soccer has gained recognition as a valid area of research itself, with attention being paid to increasingly complex skills involved in the game including the throw-in (Messier and Brody, 1986; Kollath and Schwirtz, 1988), heading (Townend, 1988), and kicking (Roberts et al. 1974; Isokawa and Lees, 1988; Rodano and Tavana 1993).

Reilly and Thomas (1976) reported jogging (36.8% of distance covered) and walking (24.8%) as the dominant exercise activities in the old English first division; cruising and sprinting accounted for 20.5% and 11.2% respectively. They also included sidewards and backwards movements, which accounted for 6.7% of the total distance covered. Average time standing still was 142 seconds, which accounted for 2.6% of the total game time, with mean duration of each rest being 3.2 seconds. Yamanaka et al. (1988) showed higher levels of walking/jogging in Japanese players (83 - 88%) but lower levels for higher intensity work (run/sprint accounting for 7 - 10%). Differences in exercise category definitions, environmental conditions, and soccer tactical strategies may account for the variability in the results of these two studies. However, despite differences in data collection method, mean total distances covered by outfield players appear to be approximately 10km (Reilly, 1996). Some studies such as Bangsbo et al. (1991) have managed to assign motion to one of nine discrete movement categories. The Danish soccer players were investigated in terms of time spent at each activity rather than on the distance covered. In terms of time the mean ratio between high speed running, low speed running and standing/walking for the Danish players was 1 : 4.3 : 7.1. These data are comparable to 1:7 high intensity to low intensity (all other activities) ratio observed by Mayhew and Wenger (1985), and the 1 : 5 : 3 sprint : jog : walk ratio reported by Brooke and Knowles (1974).

However, due to the nature of soccer, each player performs the majority of the 90 minutes playing period without the ball. Reilly and Thomas (1976) found outfield players in possession of the ball for 1.73 % of the total distance covered. Withers et al. (1982) showed a more comprehensive breakdown of skills and their frequencies, stating that approximately half of the mean total 51.4 ± 11.4 (S.D) ball contacts are with the foot (26.1 ± 12). An estimation of 25 contacts from the thigh, chest and head can then be made. Direct frequency counts were also made for non-linear actions of turns (49.9 ± 13) and jumps (9.4 ± 6.5). Direct contact with the ball also leads to alternative energy demands. Reilly and Bull (1984) showed the added energy cost of dribbling a ball on a treadmill to be constant at 5.2 $kJ.min^{-1}$, with such increases attributable to alterations in stride length and frequency from those naturally selected at a given speed. Reilly and Bowen (1984) examined additional energy costs of unorthodox movements eg. sideways and backwards. The energy cost increased linearly with speed of motion for each of the directional modes. Both running backwards and sideways elicited similar perceived exertion, whilst running forwards elevated the perceived exertion 2.3, and 4 units of the Borg Scale with increasing speeds. The quantification of the energy costs of these backwards and sideways movements implies researchers deem non-linear motion to be of importance in soccer, yet the modalities studied have often been those with a ball. Relative distances and durations of curvilinear motion remain absent from the literature, as does the subsequent analysis of these movements.

The limited number of biomechanical studies that have focused on soccer technique have generally centred on kicking. For example, Elliot et al., (1980) described the development of a mature kicking action in six characteristic phases. The effect of angle of approach to a soccer kick was investigated by Isokawa and Lees (1988) in a cinematographic study, with the authors recommending an optimum approach angle of around 30 degrees. Ball velocity from a soccer kick depends on the effective striking mass of the foot, and Plagenhoef (1977) related the effective striking mass of the leg to the stiffness of the leg and foot complex. Asami and Nolte (1983) found that while the change in ankle joint angle did not correlate with ball velocity, the change in angle at the metatarsal-phalangeal joint correlated highly significantly with ball velocity with greater deformation decreasing ball velocity. The authors proposed a reduction in foot deformation for powerful ball kicking. More recent studies Rodano and Tavana (1993) have examined the soccer kick in three dimensions, following early work by Griffiths (1984) who performed a 3D study of the female soccer kick. Alongside kinematic studies of kicking, kinetic studies are also evident in the literature. Zernicke and Roberts (1978) calculated joint forces during kicking to support a 'whiplash and flail' effect of the lower leg as energy is transferred through the knee joint. The rapid deceleration of the upper leg increasing the velocity of the lower leg towards impact. Other studies have contradicted such postulations and suggested the decrease in angular velocity of the thigh during the latter half of the kicking motion does not serve to increase angular velocity of the shank. Furthermore, that the decrease in the angular velocity of the thigh occurs as a direct result of the influence of the shank's angular motion on the thigh (Putnam, 1983). Such conflicting literature suggests further work is still required to investigate the methods of kinetic transfer during kicking, yet is beyond the scope of this thesis.

Studies of the game of soccer cover numerous and diverse topics. As basic demands of the game were assessed by both work rate and physiological measurements, investigations have become more focused, employing biomechanical techniques to assess the technical aspects of performance. The increasing demands on the players in the modern game means that research should continue to specialise in this area to improve further technique, equipment and training knowledge. In the main, biomechanical studies have focused on kicking as it is the most important and fundamental skill involved in the game. However, other aspects of soccer specific techniques need to be investigated to increase our understanding. Overground linear and curvilinear jogging and running are fundamental skills in soccer as they are used to impliment tactical manoeuvres and gain a shooting chance at the opponents' goal, yet little research has focussed on curvilinear motion. A deeper understanding of the muscles used, the forces generated and their application at the shoe-surface interface will aid practitioners to improve technique analysis and sport specific training.

2.3

ELECTROMYOGRAPHY

Electromyography has been developing slowly in chronological terms. This development was summarised by Basmajian and DeLuca (1985) in the text 'Muscles Alive'. Electromyography has since aided sports scientists to understand muscular involvement in many modern sports. Clarys and Cabri (1993) highlighted important instrumentation and experimental points, in addition to an exhaustive review of existing applications of electromyography in sports.

Many factors affect the pattern of muscle activity during gait (eg. velocity, gradient), yet many phases of muscle action remain consistent in the literature. Differences between results can mainly be accounted for by the alternative methods used, but also by the selection of muscles by groups of investigators.

In the analysis of previous research, a common definition of a complete running cycle begins with foot contact on one side and continues until the subsequent foot contact on the same side. The transition from walking to running has been shown by Nilsson et al. (1985) to occur at about 2m/s in a healthy adult. In early work describing electromyography applied to treadmill running, Elliot and Blanksby (1979) produced a comprehensive description of major superficial muscle activity using ten subjects at discrete speeds of 2.5m/s and 3.5m/s. Although the study used exclusively female subjects, the data collected provide a baseline for many modern studies. Elliot and Blanksby (1979) synchronised both EMG and CMG (cinematography) in order to describe and quantify changes occurring during the running cycle using surface electromyography of the large muscles of the lower extremity. The authors discovered that at heel strike, the thigh, lower leg and foot were moving in the same direction as the treadmill belt. Such actions reduced the resistance to forward motion and enabled stabilisation from absorption of the downward force of the runner, to prepare for the propulsive phase. At heel strike Elliot and Blanksby (1979) noted vastus muscle activity displayed greater amplitude at 3.5 m/s than at 2.5 m/s. These data agreed with the findings of Brandell (1973) who reported data for walking and running overground using fine wire electrodes. However, only raw data were reported from the study. MacIntyre and Robertson (1987) also showed vastus medialis and lateralis activity at heel strike. Elliot and Blanksby (1979) noted biceps femoris and semimembranosus were also active at heel strike for stability. Tibialis anterior activity was recorded at heel strike, coinciding with the eccentric contraction of the muscle to aid impact force attenuation. As the gait cycle proceeded, the quadriceps muscle group showed its peak activation in the support phase during leg extension around the 'heel-off' point. The support of the body weight was deemed to cause readings to peak for those muscles.

Winter (1983) reported eccentric muscle work of the triceps surae during early stance from cinematography. These data were supported by MacIntrye and Robertson (1987) who suggested the activity served to slow forward motion of the tibia during stance. The propulsive phase of the cycle was reported by Elliot and Blanksby (1979) to give high values of triceps surae activity. Such values would support the role of the triceps surae as the prime movers in this phase of the cycle. As hip extension was also involved during the propulsive phase, the authors also reported peak biceps femoris activity at that point. As hip movement continues to the limit of hyperextension, it became evident from the data presented that eccentric rectus femoris activity was not of a significantly high enough level to be responsible for the control of hip extension. However, the exact muscles responsible for eccentric control could not be identified as the authors did not measure activity in the iliopsoas. In the early phases of swing Winter (1983) and MacIntyre and Robertson (1987) noted quadriceps eccentric activity from kinematic analysis to prevent excessive flexion of the lower leg and foot. As the leg reached the limit of the backswing, hamstring activity was high, with leg flexion. Fast hip flexion then occurred coupled with knee extension. EMG data and kinetic analysis showed that knee extension was initiated by a transfer of inertial force and was then continued by muscular action of the quadriceps (Elliot and Blanksby, 1979). Such results would also agree with the data collected by Dillman (1970) and Brandell (1973) for overground running. Hamstring muscles are involved in the deceleration of the thigh and reversal of its direction prior to heel strike.

Schwab et al. (1983) found two consistently identifiable phases of quadriceps activity at 4.57 ±0.6 m/s during treadmill and overground running. The onset of the first occurred during midswing, at an average of 55% of the gait cycle, whilst the second burst of activity occurred during terminal swing, at 86% of the cycle. The mean onset of hamstring activity during midswing was at 58% of the cycle, with other hamstring activity at 2% of the cycle, just after contact. Gastrocnemius medial head was found to activate just before initial contact, at 98% of the cycle. Midswing gastrocnemius activity also consistently appeared in the investigation. Midswing calf activity was also reported by Mann and Hagy (1980) from their study on treadmill running.

It should be noted that during walking no consistent action of the quadriceps is noted in the midswing. However, during running a burst at approximately 55% of the cycle was reported (Schwab et al., 1983). Such action becomes explicable when one considers that the lower leg is extended primarily from a transfer of momentum from proximal to distal segments during walking, but reduced cycle time in running requires additional quadriceps action to extend the lower leg for initial contact (Schwab et al., 1983). As previously stated, the transition from walking to running has been shown by Nilsson et al. (1985) to occur at about 2m/s in a healthy adult.

Kameyama et al. (1990) investigated EMG patterns around the ankle joint for running in Japanese subjects. Unfortunately the authors only presented results of the gastrocnemius and tibialis anterior muscles, and reported different patterns for tibialis anterior of double burst, triple burst and continuous firing, with double burst activity being most common.

Velocity effects were investigated by Elliot and Blanksby (1979) who showed EMG activity in all monitored muscles to increase with increasing velocity of gait (2.5 and 3.5 m/s). However, Ito et al. (1985) measured the same quadriceps and hamstring muscles and showed average integrated electromyography (IEMG) for four subjects summed from six lower extremity muscles remaining constant for the support phase over the range 3.7 - 9.3 m/s, but to increase for non-support. The authors claimed that increased elastic energy during the latter support phase could explain the lack of increase in average IEMG at higher speeds.

No studies could be located within the published literature to date on the muscular activity during non-linear motion apart from one study arising from the research in this thesis (Smith et al., 1997). However, in a kinematic analysis of cutting movements, Andrews et al. (1977, cited by Schot et al., 1995) speculated that direction change is accomplished in the main through torque generated by the torso, pelvis and lower extremity musculature being applied to the ground. To ascertain whether such speculation was correct would require electromyographic investigation during non-linear activity.

Studies investigating lower extremity muscle activity in running gait have been centred on the analysis of major superficial muscles during straight overground (e.g. Brandell, 1973) or treadmill running (e.g. Elliot and Blanksby, 1979). Overall, studies report agreement in temporal firing patterns of the major muscles investigated. However, with analysed movements in running limited to linear motion, it is impossible to infer similar muscular requirements in non-linear motion, with the predominant involvement of the same superficial muscles remaining similarly unclear. Further experimental work is needed if the mechanisms of progression in non-linear motion are to be understood. Pilot testing of the major superficial muscles of the lower extremity is required during curvilinear motion to ascertain those muscles which control movement outside of the sagittal plane. The muscles chosen must therefore control the hip or ankle joints, and also not display any cross talk from adjacent muscles that perform an alternative function.

2.3.1

LIMITATIONS

Previous studies have documented possible inaccuracies in evaluating the EMG activity of certain leg muscles using surface electrodes, especially when other muscles in reasonable proximity have antagonistic function (Perry et al., 1981). Tomaro and Burdett (1993) in their study into effects of orthotics on lower extremity muscle action suggested data from the surface electrode placed over the muscle belly of the peroneus longus appeared to be receiving extraneous activity from both the tibialis anterior and gastrocnemius muscles and showed no consistent pattern of activity throughout the group of subjects. Schwab et al. (1983) also suggested that surface electrodes be used only for regional synergistic muscle groups and should not be extrapolated to compare functions of individual muscles within those groups.

Some success has been reported in the distinction of rectus femoris, vastus lateralis, and vastus medialis muscles by Nilsson et al. (1985) who studied seven lower extremity muscles with surface electrodes during gait at different velocities. Nilsson et al. (1985) verified the results from surface electromyography using bipolar intramuscular electrodes in one subject, with no consistent differences in onset time and duration of EMG activity reported in the three superficial quadricep muscles, gastrocnemius lateralis, and tibialis anterior. MacIntyre and Robertson (1987) also disagreed that surface EMG could only report wholistic muscle group activity, as they showed different muscle patterns for the vasti muscles compared to the rectus femoris during running using surface electrodes. These differences were attributed to the two joint nature of the rectus femoris, which controls hip flexion in addition to knee extension. Cabri et al. (1987) also investigated individual muscles around the ankle joint in females using surface electrodes with apparent success, although presented results were minimal. Comparisons between surface and indwelling electrodes have shown no significant differences in time of onset of muscle action (Schwab et al., 1983; Nilsson et al., 1985), and it would appear that the choice of surface as opposed to indwelling electrodes in the majority of studies was due to subject comfort and recruitment, in addition to financial and ethical considerations.

The majority of researchers have reported single-day EMG scores on a subject, usually single stride scores within a day (Yang, 1985). Such reports would imply that gait is repeatable both within and between days. Although this assumption is likely to be valid for temporal and kinematic measures, it remains to be substantiated for kinetic and EMG measures (Yang, 1985). Authors such as Elliot and Blanksby (1976) have suggested that EMG patterns are repeatable and adequately represented by an average of five strides. Yang (1985) showed that some subjects had highly repeatable EMG patterns between days for walking at a comfortable cadence, with electrode sites marked with silver chloride to ensure exact attachment. Other subjects suggested a complete different balance of synergist muscle effect to achieve a consistent temporal and kinematic pattern. Such results would lead investigators to interpret EMG results with caution, as the EMG pattern observed on any one day may be one of many 'normal' patterns of that subject's gait.

Electrode placement has been an area of discussion for many authors. Kramer et al. (1972) found a 25% decrease in the amplitude of rectified EMG (REMG) 3cm from the assumed midpoint of the biceps brachii muscle. Zuniga et al. (1969) reported no apparent advantage in placing the electrode on the motor point of the muscle as the optimal electrode location. Clarys and Cabri (1993) suggested points of electrode application for electromyographical study include :-

1) Over the motor point of the muscle
2) Equidistant from the motor point
3) Near the motor point
4) On the mid-point of the muscle belly
5) On the visual part of the muscle belly
6) At standard distances between osteologic (anthropometrical landmark) reference points.

After consideration of previous research, they concluded that surface electrodes should be positioned at the visual midpoint of the contracted muscle. Such recommendation helps in the standardisation of future EMG work, enabling valid comparison of results between investigations.

Standard methods of electrode placement must be used with all monitored muscles. To compare results from different trials both within and between subjects, however, raises issues of normalisation of the EMG signal. Usual practice has involved expressing the raw or enveloped trace as a percentage of the maximal voluntary contraction (MVC) of that muscle, or group of muscles, or of the highest EMG value obtained. More recently the normalisation topic has received attention as some investigators reported maximum values in excess of the MVC, especially during dynamic activities. Clarys and Cabri (1993) reported other techniques included normalisation to the highest peak activity in dynamic conditions, and cited a full discussion of the topic by Yang and Winter (1984).

Another important feature of EMG is that of the time lags between the onset of electrical activity and tension in the muscle (electromechanical delay, EMD). EMD can offer an explanation of the discrepancy between EMG activity and body segment motion. Komi and Cavanagh (1977) observed that the delay was shorter during eccentric contractions in human skeletal muscle when a MVC was performed subsequent to passive arm movement. Such evidence suggested that the difference was related to the rate of change of length of the series elastic element of the muscle. Norman and Komi (1979) showed that a speed of movement effect was also evident as EMD was shorter in fast as opposed to slow movements. They also suggested a relationship between muscle fibre type and EMD, but could not offer a satisfactory explanation of the phenomenon. With triceps brachii EMD being shorter than biceps brachii EMD it would appear that such a suggestion may be correct. Triceps brachii muscle has been shown (Johnson et al., 1973 cited by Norman and Komi, 1979) to consist of a greater percentage of fast twitch fibres than the biceps brachii muscle. However, the proportion of subjects for which this statement was correct was not highlighted. Norman and Komi (1979) showed EMD in both biceps brachii and triceps brachii to be between 25 and 45ms, a value which was in agreement with previous research. The EMD in the measurement of EMG ought to be considered in interpretation of results, especially when attempting to synchronise movement and EMG data.

Several contradictions have become evident in the literature of EMG in running based activities, specifically the debate over the use of surface versus intramuscular electrodes for the identification of discrete muscle activity (Schwab et al., 1983; Nilsson et al., 1985). Also, there has been a difference of opinion on the level of muscular activity with increasing velocity (Elliot and Blanksby, 1979; Ito et al., 1985). The interpretation of results emanating from treadmill and overground studies appears to have been answered, in part, by Schwab et al. (1983) who found EMG traces to be very similar for treadmill and overground running. Overall however, the majority of studies (Elliot and Blanksby, 1979; Schwab et al., 1983; Nilsson et al., 1985) go someway towards quantifying a general cyclical muscular activity pattern during gait.

The adaptive mechanisms involved in the performance of curvilinear and non-linear walking, jogging and running have not been investigated. Limitations have been highlighted for research in a sports environment but electromyography can be used effectively and accurately if these are taken into account (Clarys and Cabri, 1993). Research on muscle activity in gait has centred on motion in a straight path at designated velocities, yet mostly using treadmill running (Elliot and Blanksby, 1979; Nilsson et al, 1985) whilst soccer entails overground running. Little overground straight running has been reported in the literature, with the exception of Mann and Hagy (1980); Pare et al. (1981). For a more specific understanding of activity in soccer players and other field games, investigation of the adaptive muscular mechanisms which enable non-linear motion is required.

2.4

KINEMATICS OF RUNNING

Kinematics is defined as the branch of biomechanics dealing with the motion of points or bodies in time and space without regard to the forces that create that motion (Cavanagh, 1990). Kinematic analysis of sports actions can therefore represent a descriptive picture of what occurs during sporting activity. With differences that become evident from such analysis, kinematics can then form the basis of objective justification for the altered mechanisms used in various sporting techniques.

With reference to soccer, many kinematic investigations have been concerned with specific soccer skills, especially kicking (Isokawa and Lees, 1988; Rodano et al., 1988; Rodano and Tavana, 1993). However, although justification for this thesis emanates from the requirement to understand and gain knowledge of soccer movements, the analysis of specific soccer ball skills are of less interest here, and therefore will not be analysed within a review of relevant literature. The literature concerned with non-linear motion *per se* remains scarce, especially with relevance to soccer. However, to enable differences in linear and non-linear gait performance to be understood, kinematic data collected during straight running must be used as a starting point. Many studies (Nelson and Osterhoudt, 1971; Nelson and Gregor, 1976; Bates et al., 1978; Elliot and Blanksby, 1979; Mann and Hagy, 1980; Nelson et al., 1982; Clarke et al., 1983) have investigated kinematics during linear motion, with only a few (Greene and McMahon, 1979; Stoner and Ben-Siri, 1979; Greene, 1985; Hamill et al., 1987) reporting motion which is curvilinear in nature. A number of review articles have also dealt with the topic of lower extremity kinematics in running, including Dillman (1975), Williams (1985) and Cavanagh (1990). The present review concentrates on studies covering linear motion in running and compares those data with the limited information arising from concerning curvilinear motion research.

Dillman (1975) presented a review of the kinematics of running, which gave good coverage of the literature. However, much of the data presented have since been replaced by studies using more reliable and accurate equipment or improved experimental methods. One important issue covered by Dillman (1975), still pertinent, was the comparison of data between studies of overground and treadmill running studies. Nelson et al. (1972) studied differences between the two methods and reported that treadmill running had a longer support phase, with lower vertical and horizontal velocity of the body. Williams (1985) stated that when significant differences between treadmill and overground running had been reported, they have generally been at speeds greater than 5 m/s. Nelson et al. (1972) showed that above 5m/s stride length was increased and stride rate decreased during treadmill running, whilst Elliot and Blanksby (1976, cited by Williams, 1985) reported a decrease in stride length and increased stride rate. These two studies also reported anomalies in support time. In his review, Williams (1985) noted that the majority of studies had reported non-significant differences between treadmill and overground running. Winter (1980, cited by Williams, 1985) hypothesised that differences could be due to transfer of energy when the athlete contacts the treadmill belt, as the athlete absorbs energy from the slight reduction in speed of the belt. Winter (1980, cited by Williams, 1985) claimed that such energy change could be the cause of some of the kinematic changes observed. If however, the treadmill belt was powerful enough, the only differences between treadmill and overground running should be due to air resistance and perceptual alterations linked to the lack of horizontal progression, which will cause an adaptation of posture.

The kinematics of linear running have been described by various authors (Elliot and Blanksby, 1979; Clarke et al., 1983) and have centred on differing aspects of technique. One of the fundamental alterations that occurs during straight running at different speeds is the interrelationship between stride length and stride frequency. Running velocity is the product of stride length and stride frequency, as reported by Dillman (1975), where a stride is defined as the period of time from the occurrence of one event until that same event is repeated. For speeds up to 7 m/s, the increases in length and frequency are reported to be mostly linear, but at higher speeds typically smaller increases in stride length and greater increases in stride frequency (Dillman, 1975; Williams, 1985). When considering individual differences in stride length, one would expect a relationship to exist with body size. Williams (1985) reported conflicting results for correlations between height, leg length and stride length. Such conflicting results emphasise the individual differences between these parameters, therefore caution should be exercised in applying any predictive equations for optimal stride length based on leg lengths and velocities.

Dillman (1975) referred to the effect of training on stride length in running by citing Cavanagh et al. (1977) who compared the stride length of elite distance runners with those of good collegiate runners at 5 m/s, and found that better runners had longer absolute and relative stride lengths. However, in a four year prospective study Nelson and Gregor (1976) found that 9 out of 10 of a group of collegiate runners decreased their stride length with training. With improved performance time during the same period the authors attributed improvement to a reduced stride length. However, over a four year period there could be a myriad of other factors including increased strength and flexibility that caused increased performance, and correlation does not infer a cause and effect relationship.

As stride length and frequency have been shown to vary as running speed is altered, it follows that support and non-support times may also be affected. The evidence shows that support and non-support times decrease as running speed increases (Nelson and Osterhoudt, 1971; Nelson et al., 1972). Nelson et al., (1972) reported that support time decreased from 68% cycle time at 3.35 m/s to 54% at 6.4 m/s, with an associated increase in non-support time across the range 2.5 - 6.4 m/s. However, Nelson and Osterhoudt (1971) showed only slight differences in non-support times as speed increased with the longest times occurring at intermediate speeds. The longest non-support times would be expected to occur just before the transition from running to sprinting.

Research into kinematic analysis of running has generally occurred in the sagittal plane. Holden et al. (1985, cited by Williams, 1985) investigated foot movements outside of the sagittal plane and showed a mean value of 6.1° of abduction. The authors reported considerable variation among subjects, with abduction increasing as speed increased for almost half the subjects. Movements at the foot and ankle have been investigated in more detail to assess the pronation and supination movements of the rearfoot in the frontal plane.

At first contact with the ground, the foot has shown to be supinated 4 - 12° and following foot strike pronation occurs very rapidly (Bates et al., 1978). Maximum pronation values have been found ranging from -4 to -25° for individual runners in shoes, with mean values typically in the range of -8 to -17° (Bates et al., 1978). The velocity at which these pronation values were achieved was reported by Clarke et al. (1983) to average - 532 °/s (range -206 to -1005) for ten subjects at 3.8 m/s. Once maximum pronation has been achieved, the foot begins to slowly supinate as the heel lifts off the ground. There appears no consistent variation in rearfoot parameters associated with running speed (Williams, 1985). It may be that increased pronation is prevented by the observed changes in lower extremity kinematics with increased speed. Research is needed to provide links that such adaptation occurs to strengthen the causal links between excessive pronation and injury proposed by some authors (Bates et al., 1978; James et al., 1978; Clarke et al., 1983).

Heel lift has also been shown to have an effect on the kinematics of running. Dixon and Kerwin (1998) reported changes in ankle angle between rearfoot, midfoot and forefoot strikers when heel lift was increased. Midfoot strikers tended to show lower ankle angles with increased heel lift. In addition, rearfoot and midfoot strikers demonstrated significant increases in maximum Achilles tendon force with increased heel lift, whereas a forefoot striker demonstrated no changes in maximum Achilles tendon force with heel lift manipulation. Such results will have implications for the style of footwear selected for soccer participation, in conjunction with various footstrike classification.

Further investigation into motion occurring around the foot and ankle was performed by Viale et al. (1997) who investigated foot orientation and lower limb kinematics in running. Abduction of the forefoot was related to ankle dorsiflexion and plantar flexion velocities. Viale et al. (1997) suggested that foot lever arms used at these velocities were all long, as the last point of contact was the first metatarsal head. If the last point of contact was the 2nd and 5th heads, then a shorter, transverse axis described as a 'high gear' by Bojsen-Moller (1978, cited by Viale et al., 1997) could have been used. The Bosjen-Moller theory stated that the foot lever arm is modified by the orientation of the forefoot during the push-off phase. Because the use of different lever arms influence the mechanical action of the ankle extensor muscles, a relationship between forefoot orientation and foot-leg kinematics could be expected. Viale et al. (1997) found an inverse relationship between forefoot angle and ankle velocity, and stated this could be affected by individual differences in the muscular control of foot stiffness. As the ankle muscles also have an abduction-adduction effect on the foot, their level of co-activation could influence foot orientations. Also high muscular activity increases ankle stiffness and provides favourable conditions for the stretch-shortening cycle behaviour of ankle extensors and for fast joint motion.

Applications of kinematic analysis techniques to non-linear motion remain scarce. Some researchers have investigated non-linear movement patterns in racket sports and basketball (Lafortune, 1997). However, to understand the mechanisms of progression, and the differences of non-linear motion to that of straight running, an analysis of curvilinear motion must first be performed. Greene and McMahon (1979) published research concerning running in a circular path. Twelve male subjects were filmed at 120 frames/s running at maximum speed along a straight track and then in circular arcs of 80, 62, 36, 20, and 12 feet wearing spiked baseball shoes on a natural grass surface. Top speed, ground contact time and ballistic air time were reported to change dramatically with radius. Greene and McMahon (1979) noted that neither step length or frequency altered appreciably as a function of the radius and therefore were assumed constant for each subject, but data on these stride kinematics were not supplied to verify these claims. The authors derived an equation to predict the speed versus the radius of the curve, using inputs of the top speed of the runner and the acceleration due to gravity:

$$R(v) = v^{5/2}/g\sqrt{vmax - v}$$

Using the equation as a predictor for velocity, radius, ground contact time and ballistic air time, Greene and McMahon (1979) suggested that for 200m races the track should in fact be circular and not oval, to give a 3% faster time. Such results were derived from the prediction equations, yet with many assumptions underlying these solutions such a result should be treated with caution. Jain (1980) further highlighted the problems with oval tracks, suggesting that a discrepancy in the lanes exist, with the outside lanes being the fastest. The difference between the innermost to outermost lane was calculated at 0.07 seconds for a 200m race. Stoner and Ben-Sira (1979) cited work by Broom (1962) and Mitchel (1968) suggesting a lag-time in the order of 0.4-0.5 seconds is expected when sprinting on the curve as opposed to sprinting in a straight path. Greene and McMahon (1979) postulated the mechanisms behind curvilinear motion, yet their data at each grade of curve showed differing velocities, making comparison of kinematic variables somewhat problematic. Delecluse et al., (1998) investigated indoor track running and reported that almost all significant differences in running velocity during a 200m sprint could be accounted for by changes in stride length. Reduction in velocity when entering a curve was caused by the shortening of stride length. Once the athlete was running in the curve the stride length remained constant, while the running velocity slightly decreased as a result of normal and progressive reduction of stride frequency. When leaving the curve stride length increased again, but this did not assure an increase in running velocity, because of the further decrease in stride frequency.

In a subsequent investigation Greene (1985) performed further experiments on a concrete surface in training shoes. Thirteen more subjects were used and a wider range of radii. In addition stopwatches were used for timing rather than cinematography. Assumptions that stride frequency and stride length did not alter as a function of radius were questionable, as was the estimation of the ground reaction force variables from mathematical equations. The author claimed that the results agreed better with the theory than the original investigation, especially at radii less than 12.2m. For running at maximum speed, stride length and stride frequency were assumed not to alter, yet speed reduction with increased curve severity was deemed to occur because of the increase in foot contact time. Estimated peak vertical force was shown to decrease as the radius shortened, especially at a radius less than 11m. However, the magnitude of the reduction was only 1.89 to 1.81 *BW* relative to a change in radius of 11m to 6.1m. A reduction in vertical peak force was attributed to an increased horizontal component of the ground reaction force. An estimated increase in horizontal force was deemed necessary to create the centripetal acceleration required to proceed in a circular path. In his model, he focused on the ratio between the gravity force, and the centripetal force. The ratio was termed the Froude number, with a Froude number of 1 occurring when a runner leans at an angle of 45°. It should be noted that Greene and McMahon (1979) and Greene (1985) conducted their research with maximal velocity sprinting. Therefore, the kinematic changes suggested may not be correct for sub-maximal speeds, which remain to be investigated.

Hamill et al. (1987) investigated the effect of running the curved portion of an athletics track. Five male subjects ran at 6.31 m/s at a radius of 31.5 m whilst ground reaction force and rearfoot kinematics were monitored. Ground reaction force results will be discussed in section 2.6. Cine film was taken at 100 Hz in line with the anterior - posterior axis of the force plate and digitised four frames before and after the foot contact. Comparison was made between straight motion, inside, and outside legs of the curve. Results at touchdown showed the outside leg to be more supinated (12.76°) than the straight (7.09°), with the inside leg being already pronated (-3.98°) at touchdown. There were no differences in maximum pronation between straight and inside legs (-11.48° and -12.49° respectively), but the inside leg was -22.56°, with mean values also much greater. In considering total rearfoot motion, the outside leg showed higher values of 25.26°, compared to approximately 18° for the other two conditions. All subjects were reported as rearfoot strikers, and Hamill et al. (1987) suggested that the difference in initial rearfoot motion was due to the necessity for landing on the lateral aspect of the heel.

Stoner and Ben-Siri (1979) compared the acceleration phase of sprinting in a curved and straight path using sagittal plane kinematics. Findings showed the stride length to decrease significantly in the curvilinear trials, with a trend for increased running times. The authors postulated that the inside leg and outside leg kinematics are inherently different when running in a curved path, with the outside leg displaying the shorter stride length. They suggested results may indicate that the outside leg is in a more advantageous position for applying centripetal force during the support phase. The lack of a shorter flight time at the inside leg was explained by the increased distance of travel for the outside leg during this period. These results were recorded on the acceleration phase of a sprint however, and therefore may not be applicable to constant paced motion. However, kinematic investigation into constant paced curvilinear motion should reveal any differences in actions of the inside and outside legs and will be addressed in the following chapters.

2.4.1 LIMITATIONS

Many investigations have examined straight running at a range of velocities. Significant contributions to knowledge in the area have been reviewed, particularly by Cavanagh (1990), with baseline data for lower limb kinematics of linear running being generated. These baseline data for linear motion provide other investigators with comparative measures against which curvilinear trials could be compared, enabling the mechanisms governing curvilinear progression to be uncovered. The limited number of studies of curvilinear motion include the trials of maximal velocity by Greene and McMahon (1979) and Greene (1985), but these do not enable the differences in kinematics to be elicited due to the varying velocity at each grade of curve. The mechanisms of non-linear progression presented by Greene (1985) were speculative, and have still to be substantiated. Hamill et al. (1987) presented measurements of rearfoot motion in curvilinear motion, yet provided no information to the altered sagittal plane kinematics between linear and non-linear motion. The characteristics and mechanics of curvilinear motion have still not been fully investigated and documented. Fundamental experimental data regarding movement at discrete velocities along several paths of constant curvature would provide a basis for understanding the mechanics of curvilinear motion. If such experimental data were to be relevant to soccer, suitable soccer-specific curvature paths should be considered at relevant jogging and running speeds. Such thoughts raise a clear research question that will be addressed in the following chapters.

2.5

SHOE-SURFACE INTERFACE

Good ergonomic practice requires the optimisation of the interface between the athlete and the sports surface. Such interaction concerns both the sports surface, and the athletic footwear. If either of these characteristics were changed, the nature of the shoe-surface interaction would also be altered. However, most research has tended to centre on one or other aspect of the interaction. The following sections are therefore divided into footwear considerations, followed by surface considerations.

2.5.1
FOOTWEAR CONSIDERATIONS

The interface between the sports performer, the sports shoe and the sporting surface has received considerable attention from researchers in the past two decades (Cavanagh, 1990). The vast majority of the work however, was directed towards the development of sports footwear in an attempt to improve athletes' performance, with much work centred on the reduction of excessive subtalar joint pronation during gait.

Pronation naturally occurs during the support phase to dissipate shock and compensate for terrain, and is essential for the transfer of weight from one foot to the other. At heel strike the tibia is internally rotated and the hindfoot becomes everted. The subtalar joint translates the internal rotation into eversion of the calcaneus. With the hindfoot everted the subtalar joint unlocks the midtarsal joints and produces a parallel configuration allowing pronation of the foot (Andelaar, 1986).

The pronation motion is passively governed by the position of the hindfoot, mobility of the subtalar joint, integrity of the supporting ligaments, and configuration of the joint surfaces of the midfoot (Andelaar, 1986). In the propulsive phase supination of the subtalar joint is coupled with inversion of the calcaneus and external rotation of the tibia. Such action stabilises the foot and enables it to act as a rigid lever for propulsion (Gross and Napoli, 1993).

Increased levels of calcaneal inversion and/or eversion was shown by Luethi et al. (1986) to correlate positively with the injuries in tennis, where many sidestep manoeuvres are performed. However, as correlation does not infer cause, such results should be treated with caution. Excessive pronation motion which continues past mid-stance can prevent the conversion of the foot into a rigid lever for propulsion and imposes functional limitations upon the joint structures of the lower limb as the foot prepares to leave the ground (Tiberio, 1987 cited by McCulloch et al., 1993).

Running shod rather than barefoot will cost the athlete 3-5% more energy, partly due to the added mass (Segesser and Nigg, 1993). Every additional 100g mass added to the foot will add approximately 1% to energy requirements (Frederick et al., 1982). Sole stiffness itself may also add to the energetic cost of locomotion, as vertical movement of the mass centre increases with decreasing sole stiffness, giving an increase of 1-2% for centre of mass movement of 5 mm (Segesser and Nigg, 1993). The shoe aims to provide the runner with added stability, grip and cushioning properties to enhance performance. The cushioning properties of the sports shoes themselves may also have an effect on the gait of the performer. Cushioning elements affect the stability of the calcaneus and an incorrect alignment of the stabilising heel cap may mechanically irritate the Achilles tendon in push off (Segesser and Nigg, 1993). In the literature the topic of cushioning has mainly been considered in relation to external forces, with the associated mechanical and biological effects of such forces (Segesser and Nigg, 1993).

Over the years many manufactures have produced technical advancements in their running shoe ranges, some of which have also been adopted for soccer footwear. Many modifications have been associated with the stabilisation of the foot in attempt to reduce overuse injury. In the soccer shoe plastic heel counters attempt to cradle the calcaneus and minimise movement at the subtalar joint, which effectively limits excessive pronation (Craton and McKenzie, 1993). However the limit of pronation angle does not assure good rearfoot control (Ferrandis et al., 1992). Such control can be further reduced in soccer due to soft surfaces combined with the undulating nature of natural turf. Stacoff and Kaelin (1989) reported that pronation occurs not just at the heel, but up to 60% along the foot and suggested that a larger rearfoot bedding would be required to control excessive pronation. The inclusion of other structures in the upper vamp can also effectively reduce the amount of pronation without increasing the torsional stiffness of the shoe or boot. Such inclusions would include posts built into the internal rear part of the shoe, and the inclusion of backers for the lacing (Ferrandis et al., 1992).

The cut of the last also affects the movement of the foot. The straighter the last has been cut, the less pronation would be permitted. Alternatively a rigid cavus foot would utilise a curved last to encourage pronation, with the last being a slip last rather than a board last to enable torsional movement. Such advancements in shoe technology enable suitable shoes to be bought without the need for orthotic intervention.

Although there are more complex movement patterns involved in soccer, the findings of running studies should still be applicable as soccer is a running based sport. Clarke et al. (1983) investigated the relationship of the features of shoe design and rearfoot control, altering sole hardness, heel height and lift, and the angle of sole flare in running shoes. They reported soles softer than 35 Shore A durometer allow significantly more pronation due to deformation of the sole by the calcaneous. Nigg

(1986) found lower pronation with 25 Shore A durometer but explained his findings by reference to shoe construction techniques. However, according to Kaelin et al. (1985) hard soles give an increased lever on impact compared to soft soles, suggesting possible increased pronation velocities. Nigg (1988) reported that earlier results showed that the shoe hardness of the total midsole could be changed without an influence on the impact forces. Such a suggestion would infer that any difference in sole hardness was to be accounted for by the body modifying motor programs to maintain ground reaction force within reasonable limits. De Wit et al. (1995) reported that hard shoes showed smaller impact force peaks at touchdown, with soft soles demonstrating larger maximum pronation and eversion angles and therefore loading the inverting muscles. Such results indicate further the capacity of the body to adapt movement patterns to perceived differences in cushioning. Cushioning is important to protect from chronic overloads in running activities, but we cannot alter the loading on the biological structures if the change in load is not perceived. Such perception of cushioning was investigated by Hennig et al. (1996) using pressure and ground reaction force measurements. Forces and pressures under the foot increased with reduced perceived sole hardness. Hennig et al. (1996) showed that with harder soles, subjects altered loading patterns under the feet to lower impact forces and increase weight bearing on the forefoot structures.

Other shoe design features investigated include lateral flare on the shoe sole, thought not to be beneficial as it could act as a lever arm that actually increases rearfoot angular velocity (Cavanagh, 1981). Heel flare has limited practical application to soccer footwear due the different movement patterns, and its incorporation could lead to buckling in certain situations. Clarke et al. (1983) concluded that shoes softer than 25 Shore A durometer having less flare will allow significantly more pronation and rearfoot movement. The authors also concluded that pronation velocity would increase as shoes became softer, having less flare and heel lift.

The inclusion of impact cushioning devices in the midsole increases shoe heel height. Clarke et al. (1983) found no significant effect of heel height on pronation, findings contrary to those of Bates et al. (1979). Stacoff and Kaelin (1989) reported that pronation was reduced for heel heights 2.3 < height <3.3cm but was increased both above and below that range. The reported reduction was explained by a slight decrease in lever arm length with higher heel height and a constant sole width.

Many soccer footwear modifications came as a result of running research. Running (heel-toe) actions are very common in soccer (Reilly and Thomas, 1976; Yamanaka et al., 1988). However, the movements in running are cyclical, in contrast to soccer, when non-linear and lateral movements are frequently executed. Very often players are forced to slow down in one sudden step eg. kicking, cutting, turning. The athlete should be protected to some extent by soccer footwear, as such movements are possibly the cause of micro traumas in ligaments, muscles and bones of the ankle and foot.

One of the primary differences between soccer and running footwear is the sole unit. Soccer footwear contains either moulded or studded cleats, in various numbers and orientations. The effect this may have on athlete behaviour during running remains unexamined in the literature. However, Lewis and Grimshaw (1993) investigated the differing kinematic patterns when running in spiked athletic shoes and flat-soled training shoes on a synthetic track. Plantar flexion during stance was significantly increased in the spiked shoes, whilst qualitative data showed a more pronounced mid-forefoot landing in spikes and a more upright trunk position. The data were collected during straight running, and therefore does not infer a similar relationship during non-linear motion.

Soccer footwear has different characteristics to those of running shoes due to different movement patterns. When landing from a lateral movement, a thicker sole would extend the lever arm to the subtalar joint axis and thus increase the supinatory movement and the likelihood of buckling. Such a situation can be seen in Figure 2.1 where the point of application of the ground reaction force occurs earlier and further from the subtalar joint as the thickness of the sole increases. Such a situation creates a greater moment arm with respect to the subtalar joint axis with increased sole thickness (right hand side).

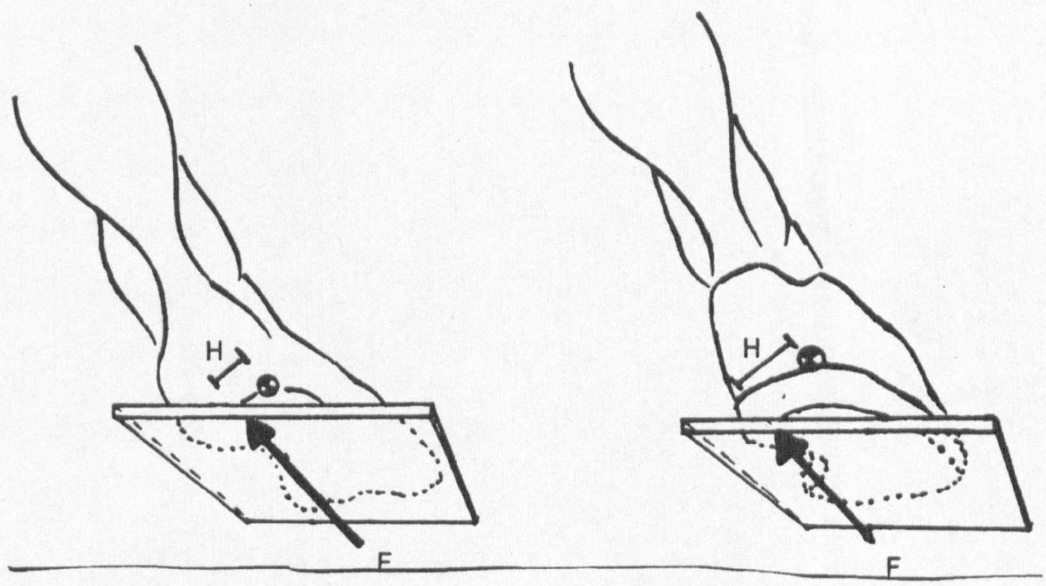

Figure 2.1 The effect of sole thickness on the moment arm of the subtalar joint axis when landing from a lateral movement.

For this reason soccer boots have flatter soles to decrease the load on the muscles and ligaments during lateral movements. A similar logic is applied in the choice of smaller studs in harder surface conditions to reduce the lever arm of the subtalar joint axis, in addition to distribution of in-shoe pressure over a wider surface area. Some soccer footwear has included midsole devices for cushioning, however heel height is not as large as in running shoes due to the nature of the more compliant turf surface.

Early biomechanical investigation into soccer footwear by Johnson et al. (1976) compared the ligamentous load of high and low cut soccer boots. The authors concluded that the old high ankle boot was 50% stiffer than the low cut boot. Also, if a low cut boot were to be worn, it should be of a soft material to reduce ligamentous strain. A high boot was deemed to offer more protection but may not be preferred for mobility reasons.

Investigation into field games' footwear has since centred on the findings of shoe research from American football, e.g. Torg et al. (1974), who investigated sole design during football specific movement tasks. The authors attempted to classify various shoe designs as 'safe' or 'not safe,' dependent on the results of several measurements of translational and rotational friction using mechanical tests. These results were then related to American football knee injuries during a season. The footwear demonstrating the lowest values of rotational friction gave the lowest frequency of injuries. It must be remembered, however, that a reduction in injury prevalence through lower friction could also result in slower movement patterns and a subsequently poorer performance, yet this was not investigated. However, soccer comprises different movements to American football, such as dribbling and cutting which occur frequently. In such movements, the high frictional forces exerted between shoe and surface may produce excessive force on the players' locomotor system, and hence, the shoe-surface relation may be more critical in soccer than American football (Ekstrand and Nigg, 1989).

Bostingl et al. (1975) measured torques using various types of shoes on different playing surfaces, including natural turf. Torques were measured using a strain gauge whilst the natural turf was securely encased in wooden pallets. The turf used was approximately three years old with an extensively developed root structure in approximately 3 cm of top soil, with grass cut to a uniform blade height of 3 - 5 cm. All testing was performed on the same day to minimise variability in moisture content, and each test performed on an unused portion of turf. Results showed that torque increased as player weight increased, with 70% more torque produced when the whole foot was in contact with the ground compared to only the ball of the foot. Such results gave a strong correlation between the number of studs (surface area) in contact with the turf and the torque values produced.

To assess the role of cleat (stud) design as a possible risk factor to anterior cruciate ligament (ACL) injury, Lambson et al. (1996) investigated the incidence of cruciate damage wearing four styles of sole configuration. The prospective study was performed over three years and showed the sole that gave higher torsional resistance when measured mechanically was associated with significantly more ACL injuries (0.017%) than the other three sole units combined (0.005%). The design was referred to as the Edge (longer irregular cleats at peripheral margin of sole with smaller pointed cleats positioned interiorly). However, many knee injuries are a result of body positional vulnerability and can occur regardless of the shoe worn or the playing surface (Bostingl et al., 1975).

Design of the soccer boot has changed little over recent years. Earlier attempts to introduce a swivel football shoe by Cameron and Davis (1973) were barred by the game's governing body. The authors suggested that removing the heel cleats reduced knee injuries, but there was still a three times greater risk than the (then) new swivel shoe. More recently, Monto (1993) suggested that it was time for redesign as the modern soccer shoe provides little protection, very little support and no cushioning.

In soccer not only does the foot participate in moving the whole body, but it also provides the means by which to advance the ball tactically and to strike at the opponents' goal. Thus it should be considered to have an essential role, especially in soccer (Saggini et al., 1993). An epidemiological study was performed by Saggini et al. (1993) on 200 professional soccer players with injured feet. The authors suggested bad boot fitting and incorrect stud positioning by manufacturers resulted in many foot problems.

Nigg (1990) suggested research in soccer shoe construction should address the compromise between foot protection and stabilising properties of the soccer shoe, the desired weight of the shoe, and the range of motion in the foot and ankle with respect to optimal performance. However, it also remains important to recognise the demands of the athlete, the sport, and the materials selected, in addition to the biomechanics used to aid design of a soccer shoe for optimal performance at the shoe surface interface (Rodano, 1992).

In summary, previous research into shoe construction effects at the shoe-surface interaction arose predominantly from research into straight running. Although the conclusions shown in the literature will remain valid for heel-toe action in the soccer player, alteration of soccer boot structure has not yet been investigated for non-linear motion. Modern soccer boot construction would appear to contain many of the characteristics for optimum performance, however frictional properties of the sole have been altered by some manufacturers in recent years with the effect on non-linear gait yet to be quantified. It is necessary to monitor alterations with athlete based tests to assess interaction during linear heel-toe and non-linear soccer specific actions to determine the human kinetics at the shoe-surface interface. Such information will enable comparison of modern and traditional soccer footwear with respect to frictional, torsional, and force generation aspects.

2.5.2
SURFACE CONSIDERATIONS

The majority of research concerned with playing surfaces has not focused on soccer. Little research has been conducted on natural turf, with only slightly more attention given to artificial soccer surfaces. From the point of view of injuries and performance, cushioning and friction properties of a surface are assumed to be of importance (Nigg, 1990). Nigg (1983) also suggested that mechanical characteristics (cushioning and stiffness) of sports surfaces may be associated with sports injuries. The cushioning ability of a material can be described as its potential to reduce impact force peaks. The stiffness of a material is defined as the ratio between the force applied perpendicularly onto a surface and the corresponding deformation in the direction of the applied force. However, it has yet to be shown conclusively that frequencies or types of injuries are directly related to the stiffness characteristics of sports surfaces (Nigg, 1990).

To create standards of sports surface within which one can hope to both reduce injuries and increase performance, specific tests must be conducted. Such tests generally fall into two categories; material tests and subject tests. Tests that have attempted to quantify a surface's cushioning characteristics are considered first.

The drop test method used for the DIN test (DIN 18035), often used in Europe, belongs to the material test group. The test uses the Artificial Athlete Stuttgart (AAS) or the Artificial Athlete Berlin (AAB). With both types of equipment a mass is dropped onto the test-foot which contains sensors, and the measures recorded are the time histories of the vertical reaction force, the vertical deformation of the surface, and the loss of mechanical energy (Nigg, 1990). Other material tests involve dropping a mass where the impact sensors are mounted on the dropping mass, which gives the advantage of more adaptability as a field test. Stanitski et al. (1974) attempted to simulate head impact with the surface, and dropped a sphere of 7.5 kg from 2.2 metres onto various soccer surfaces and found natural grass showed the greatest absorption of 89%.

Subject tests for cushioning properties involve a subject performing an action typical to that executed in the relevant sport, with sensors mounted to either the soles of the feet, the shoes, the floor, or a combination. With additional information from kinematic analysis it is also possible to estimate the internal forces acting on the subject (Nigg, 1990). Another category of subject test would involve a subject performing a typical action whilst the movement of the surface is monitored using high speed filming techniques.

Whilst all types of test mentioned have advantages in certain situations, it is important to realise that many also have shortcomings. For all material tests involving the dropping of a mass onto either a test foot or surface, it must be remembered that deformations will be dependent on the mass and the drop height. Both of these must lead to an impact representative of that which would regularly be encountered in the sport. Also, inertia terms ought to be taken into account when assessing surfaces as the literature describing test procedures does not report that this has been done (Kolitzus, 1984). Material tests have the disadvantage that they report reactions to a standard impact, whereas with a real athlete, movement patterns are modified slightly on differing surfaces. Such alterations are not detected with material tests and thus may account for the missing correlation between results of peak forces obtained from material and subject tests, since the different body limb masses may change their acceleration histories on different surfaces (Nigg, 1990).

A sports surface ought to also have its properties tested in terms of frictional characteristics. There is evidence that frictional characteristics of sports surfaces are connected to surface related injuries (Torg et al., 1974; Nigg and Yeadon, 1987). The friction between the shoe and the surface can be either rotational or translational. Translational friction usually depends on the material and the structural patterns of the surface and the shoe, and is assumed to be independent of weight and surface area. The moment of rotational friction depends on the pressure distribution in the contact area and the size of that area. Tests using subjects to quantify frictional characteristics appear to show that rotational friction is maintained below a limit of about 25 Nm by modifying movement patterns to avoid higher moments. Thus, subjects may not predict frictional characteristics for movements that are not controlled (Nigg, 1990). Such a situation could lead to a player encountering high levels of rotational friction, predisposing the tissues of the lower extremity to injury. Translational friction was assessed by Stanitski et al. (1976) who attached 11.5 kg to a size 13 shoe and performed drag tests. No grain effect was shown on any of the tested surfaces, including natural grass and artificial turfs, with natural grass showing the lowest force required to initiate drag. Such results would suggest natural grass should provide least risk for surface related fixation injuries.

Inherent within the work of Hamill et al. (1987) in curved motion in athletics, was an inference into possible injuries caused by the motion. This research into athletes performing curvilinear running in athletics stated that although similar patterns in the passive ground reaction force peak were evident between the inside and outside limb, the supinated position of the outside foot of the turn at heelstrike could cause injury. The supinated foot is anatomically locked at the midtarsal joint and therefore may not be able to deal with the shock at impact (Hunt, 1985 cited by Hamill et al., 1987). The inside foot during curvilinear motion lands in a more anatomically advantageous position to dissipate shock. Hamill et al. (1987) claimed pronation angles of -22.56° for the inside leg were excessive, but necessary as 10° extra pronation is required to reach foot flat due to body lean. The increased mediolateral forces could stress medial tissues on the inside leg, yet the foot is better positioned to absorb shock. They also inferred that the inherent stresses encountered during curved running could lead to injuries. The outside leg would be more susceptible to impact injuries and injuries of the foot due to increased ranges of motion, whereas the inside leg would be more likely to suffer foot injuries due to overpronation. However, this investigation of non-linear motion in athletics examined running in only one direction, whereas non-linear motion in soccer can occur in any direction. Therefore, such a specific mechanism for injury during non-linear motion may not be as applicable to the greater array of movements which occur in soccer.

Soccer surfaces have been constructed from a variety of different materials including dirt, gravel, plastic and natural grass, although over recent years the two surfaces of natural and synthetic turf have been dominant. Traction and uniformity differences between synthetic and natural surfaces are factors which have generated extensive controversy (Krahenbuhl, 1974). Although synthetic turf has been shown to be more uniform, proponents of natural grass see non-uniformity to be an integral part of the sport (Winterbottom, 1985). A report commissioned by the Football Association (Winterbottom, 1985) proposed that top level soccer was to be played exclusively on a natural turf surface. Therefore, experimental work in soccer should be conducted natural turf at the interface for ecological validity.

An insight into what occurs at the lower extremity in soccer players during game specific non-linear actions such as cutting, turning etc. could increase our understanding of the mechanisms involved when the player performs such actions on natural grass surfaces using soccer footwear with high friction soles. Research into quantifying the effects of altered shoe-surface interface in soccer also remains an area where greater knowledge may aid performance in the modern soccer player, and forms a research question for this thesis.

As alteration of any one part of the surface or the shoe characteristics can change the nature of the shoe-surface interaction, to understand the effects of any new combination, surface and subject tests must be performed. Whilst existing surface tests may be appropriate, each separate scenario must be viewed objectively to determine which test to employ for valid results.

2.6

GROUND REACTION FORCES OF RUNNING

Early scientists and philosophers had recognised the ground reaction force to be vital in the locomotion of the human body. Cavanagh (1990) reviewed the graded historical development of ground reaction force measurement which culminated with the high performance measurement equipment now available. Today most platforms are connected on line to a micro or mini computer. With such a system, however, it is important to have a minimum of a 12-bit non-integrating analogue to digital (A/D) converter (Miller, 1990) to provide sufficient sensitivity.

Measurement of ground reaction forces (GRF) occurs in three axes; vertical (Fz), anterior-posterior (Fy), and medio-lateral (Fx), with the last two composing the horizontal forces that can be combined to give frictional measurements. Mean ground reaction force data are affected by the footstrike of a runner (Cavanagh and Lafortune, 1980) with the largest vertical forces of about 2.5 Bodyweight units (*BW*). During running the mean vertical reaction force for rearfoot strikers shows a double peaked curve, which is usually described as, first, the impact peak and second, the 'active,' 'thrust,' or 'drive-off' peak, from the actions used to produce them (see figure 2.2). Cavanagh and Lafortune (1980) investigated seventeen distance runners at 4.5 ms^{-1} and showed the first peak rising to approximately 2.2 *BW* in 23 ms. The second peak had a lower loading rate, attaining an average maximum value of 2.8 *BW*, 83 ms after initial contact.

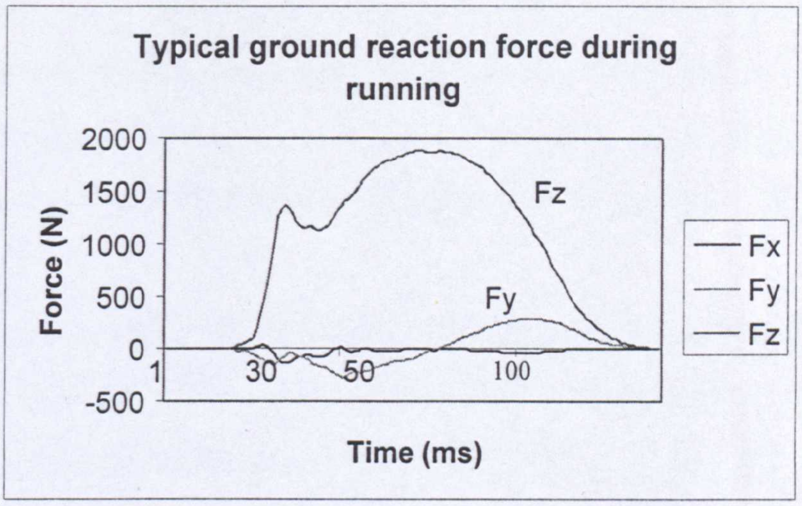

Figure 2.2 Typical ground reaction force pattern during running. (Fx - mediolateral force; Fy – anterior-posterior force; Fz – vertical force. Data from University College Chichester Laboratory)

The majority of studies undertaken investigating ground reaction forces in sport have focused on running. In the last few decades many key articles on ground reaction forces in running have been written including evaluation of running shoe characteristics (Bates et al., 1983; Lees and McCulloch, 1984; Frederick, 1986), and examination of ground reaction forces in running (Williams, 1985; Nigg, 1986; and Munro et al., 1987). As soccer is a running based game the measure of ground reaction force in running would be relevant in the analysis of soccer movements, as Reilly and Thomas (1976) reported a division in motion of 36.8% jogging, 20.5% cruising, and 11.2% sprinting during an English first division game. Other soccer specific actions such as dribbling also involve activity based on running.

Miller (1978, cited by Cavanagh and Lafortune, 1980) presented data for a slow jog which indicated vertical forces of approximately 2.1 *BW*, 25ms after ground contact. Later in the support phase the vertical component exceeded 2.5 *BW*. Nigg (1986) described the first peak after contact as the 'impact peak.' The second peak, during midsupport is referred to as the 'active peak,' indicating the role that the muscles play in its development. Cavanagh and Lafortune (1980) categorised runners according to their footstrike characteristics, with athletes classified as heel strikers, midfoot strikers or forefoot strikers. For midfoot and forefoot strikers, the impact peak is typically attenuated or absent (Cavanagh and Lafortune, 1980; Clarke et al., 1983)

Cavanagh and Lafortune (1980) showed that different footstrike characteristics produced different vertical reaction forces, with rearfoot strikers attaining an average maximum value of 2.8 *BW* (S.D. 0.3 *BW*) and midfoot strikers 2.7 *BW* (S.D. 0.2 *BW*) with an absence of an impact peak for the midfoot strikers. McKenzie et al. (1985) suggested a rearfoot strike gives a peak of around 2.9 *BW*. For the majority of studies of running speeds between 3 to 6 m/s, the vertical reaction force is usually between 2 and 3 *BW* (Miller, 1990). Cavanagh and Lafortune (1980) generated the term centre of pressure because when the whole of the shoe is in contact with the ground, the forces acting over the sole can become confusing. The use of the centre of pressure traces throughout a running support phase led to the authors devising the 'strike index' as a way of determining footstrike characteristics.

Mann and Hagy (1980) reported impact peaks of 2.5 - 3.0 *BW* at 12 miles/hour (5.33 $m.s^{-1}$). Munro et al. (1987) reported an increase from 1.6 *BW* at 3.0 m/s to 2.3 *BW* at 5 m/s which agreed with reported values by Miller (1978 cited by Cavanagh and Lafortune, 1980), Clarke et al. (1983) and Hamill et al. (1983) for running speeds of 4.0 and 5.0 m/s respectively. Vertical ground reaction forces increase with speed, and during sprinting, vertical forces can be as great as 5.5 *BW* at 7.52 m/s (Payne, 1978).

Ground reaction forces have been reported to alter dependent on whether the athlete is shod or barefoot. Stockton (1993) studied 9 male runners at 4.5 m/s both barefoot and shod, and reported GRF data showed significantly higher barefoot vertical impact force (2.23 ± 0.55 *BW* shod; 2.66 ± 0.76 barefoot), greater barefoot loading rate, and a lower barefoot minimum force. Anterior-posterior forces showed a significantly shorter barefoot time to peak braking force with significantly shorter barefoot stance time.

The anterior-posterior force is reported to be biphasic (Munro et al., 1987) because of its braking-propulsion action. It exhibits single, double, or multiple impact peaks which do not appear linked in any simple way to footstrike classification (Munro et al., 1987; Miller, 1990), as was previously believed by Cavanagh and Lafortune (1980), Hamill et al. (1983) and Payne (1983). Cavavagh and Lafortune (1980) reported that rearfoot strikers typically show a single retarding peak and midfoot strikers, biphasic peaks. They found braking maxima of 0.43 and 0.45 *BW* for rear and midfoot strikers, and propulsion forces of 0.5 BW at 4.9m/s. Payne (1978) reported braking-propulsion forces in sprinting as 0.8 *BW* and 0.6 *BW* respectively. Peak shod braking forces of 0.518 *BW* ± 0.11 and barefoot values of 0.58 ± 0.13 were reported by Stockton and Dyson (1998) at approximately 4.5 m/s. Researchers tend to agree with Cavanagh and Lafortune (1980) who reported fore-aft shear to occur at 48%, that transition from braking to propulsion occurs around or just below 50% of the stance time.

It has been possible to identify characteristics of the mediolateral force traces in running. However, data from this component have been associated with high variability. Consistent patterns have only been associated between footstrikes with the same foot in a single subject, with variability in magnitudes and number of zero line crossings associated with different subjects and foot contacts (Miller, 1990). The great variety in foot placement among individuals may be the reason for the high variability in the mediolateral ground reaction forces (Williams, 1985). Lees (1988) reported the mediolateral forces to be "more complex," and gave maximum magnitude of around 0.25 *BW*, higher than values around 0.1 to 0.2 *BW* reported by Cavanagh and Lafortune (1980) at 4.5 ms^{-1}, and Bates et al. (1983). Williams (1985) claimed a strong correlation (r=0.71) between average mediolateral impulse and position of the foot relative to the midline of progression during ground contact. Such results are logical in mechanical terms, as a greater overall force would be required to return the body centre of gravity over the line of progression in preparation for the subsequent stride. Presuming the foot was placed further from the line of progression, a greater mediolateral impulse should, therefore, be noted.

Many researchers (Clarke et al., 1983; Nigg, 1985) have used measures of ground reaction force to assess properties of running shoes. An apparent contradiction in the literature became evident when considering the effect of shoes on the ground reaction force. Nigg (1985) noted soft shoes produce a delay of the impact peak, whilst Nigg and Bahlsen (1988, cited by Miller, 1990) opposed the earlier work by showing harder midsoles displayed the lowest maximal vertical loading rate. Such results may be caused by modifications of the motor patterns, through which the athlete tries to adapt their characteristics to different cushioning systems (Rodano, 1992). In soccer, some modern boots are designed with specialist cushioning systems adapted from running shoes, which can take the form of foam, Air or Gel inserts into the midsole. These systems, as they are not required for impact attenuation on hard surfaces, tend to be accommodated in smaller midsoles, giving less heel raise than their running shoe counterparts. The majority of soccer boots do not include any impact cushioning systems, despite the dominance of heel-toe activity in the sport. They rely on the softer natural grass surface to provide reduced severity at impact by point elastic deformation. It would be suggested that manufacturers incorporate some impact cushioning system into their soccer footwear, especially for footwear designed for harder surfaces. However, care must be taken not to significantly increase the heel height and therefore the susceptibility to ankle injury from increased turning moment to the subtalar joint axis.

2.6.1

LIMITATIONS

Many studies using ground reaction force measures reported only peak forces recorded in each of the three planes. Such data reduction involved the rejection of quantities of relevant information, especially if the data was sampled at the correct frequency. Alternative strategies have been to report average forces, or average impulses from the recorded trace. The lack of standardisation in reporting of ground reaction force variables led Bates et al. (1983) to identify 20 critical variables for the comparison of the ground reaction force pattern. Measurements include temporal, force and impulse values from the three force planes. These variables are now taken as the benchmark when reporting ground reaction force during gait.

The number of trials to be sampled before valid data are achieved has been an area of conflict in the literature. Bates et al. (1983) suggested mean values of eight trials were taken to obtain stable and reliable data, though their selection of criteria for stability was subject to debate (Williams, 1985). Five trial averages have often been used (Cavanagh and Lafortune, 1980; Clarke et al., 1983) and could probably be considered sufficient. In deciding the number of trials used, the reproducibility of a given movement is critical, as is the minimum number of trials in order to maintain statistical validity.

Another measure of ground reaction force that has received relatively little attention, has been the moment about the vertical axis acting through the centre of pressure. This value is termed the free moment (Mz′). Mz′ represents the force couple resulting from friction forces between the foot and the ground and is calculated by subtracting the r (radius) x F (force) shear force from the moment about the vertical axis of the force plate (Mz). Holden and Cavanagh (1991) presented data using the free moment to assess differing amounts of pronation in a perturbation study. The authors reported the free moment acted to resist the abduction and adduction components of pronation and supination at the subtalar joint. The free moment acted to resist foot abduction during 71% of foot contact. A rotational friction coefficient can be obtained by dividing Mz′ by Fz. Such a value would appear useful for the analysis of non-linear turning movements, yet has only been documented by one author, Stuke et al. (1984), who applied the measure to calculate frictional coefficients during ninety degree turning movements. Free moment calculations may be useful in assessing the relative contribution of the inside and outside legs to the rotatory force applied to the ground during curvilinear motion. Such measurements will be considered later in the thesis.

In soccer, little investigation has evolved in quantifying GRF. As mentioned previously, the majority of activity performed in a game entails jogging and running (Reilly and Thomas, 1976). However, these ground reaction forces cannot be likened to forces measured in published scientific running papers due to differences in footwear and surface. In addition, the movements of jogging, running and sprinting which occur in soccer are not always straight, which limits the value of straight running experimental studies from electromyographical research on overground (Brandell, 1973) or treadmill running (Nilsson et al., 1985). Many other soccer specific actions of tackling and turning are unique to a game situation and have been difficult to standardise and reproduce for experimental evaluation.

Kicking represents an activity which places the shoe-surface interface under great stress (Lees and Kewley, 1988), and the forces generated are therefore of interest. Ground reaction forces in soccer kicking have been quantified by several authors, including Rodano et al. (1988) who showed that the vertical force (Fz) on the support leg during instep soccer kicking was 1.93 to 2.36 *BW*, with the horizontal forces (Fy and Fx) 0.88 to 0.5 *BW*. The study used indoor soccer footwear on an artificial (Astroturf) surface. Similar findings were also noted by Isokawa and Lees (1988) using six male soccer players at different angles of approach for maximal instep kicking. The researchers showed no difference in vertical ground reaction force at different approach angles. Rodano and Tavana (1990) showed maximum Fz values of 3.20 *BW*, and average Fz readings of 2.69 *BW* and a horizontal reading of 1.24 *BW* with data from ten professional players. Such values may appear large, yet, when landing after jumps in training shoes with the foot in a horizontal position, forces can rise to 6 *BW* (Valiant and Cavanagh , 1983 cited by Stacoff and Kaelin, 1989). The authors reported a reduction to 4.3 *BW* when landing on the toes, slightly greater than the 3.5 *BW* reported by Nigg et al. (1984). Stacoff and Kaelin (1989) suggested that landing with the foot in a horizontal position would only occur in fatigue of the triceps surae. Values for the support leg during kicking are of similar magnitude to those reported above from distance running studies, but less than those reported from vertical jumps. However, horizontal forces from professional players (Rodano and Tavana, 1990) are in excess of those from other soccer studies (Rodano et al., 1988) and from distance running (Cavanagh, 1990), due to the requirement of the final stride to stabilise the body in preparation for kicking. Such a braking action would be more pronounced in professional subjects with superior development of leg musculature.

The gait of twenty four International soccer players in addition to forty high ability players was investigated by the use of force plate and pressure measurements by Saggini et al. (1992). Vertical force maximum of 1.48 *BW* and mean of 1.26 *BW* were reported. Such figures would suggest that the subjects were walking but no mention was given to the speed of approach, surface or footwear used. Saggini and Vecchiet (1994) studied male and female soccer players running over a force platform at 2.8 m/s. The subjects were elite soccer players and were compared to a control group of normal subjects. Differences were found in the impact peak of the vertical ground reaction force, with males demonstrating significantly higher values (1.48 *BW* compared to 1.33 *BW*) than females. The authors concluded that vectograms of the male and female soccer players were essentially the same, with running tests not demonstrating any significant modifications of the ground reaction force compared with normal subjects at the defined velocity. In another study Saggini et al. (1992b) concluded that the ground reaction pattern of the professional soccer player is repetitive and typical. However, data presented were minimal and details were omitted regarding footstrike characteristics, players position and mass.

Many differing forms of non-linear motion occur in soccer, with little attention being devoted to their analysis using GRF measurement. One such movement involving an abrupt change of direction, such as an attempt to avoid or evade an opponent or obstacle, is commonly known as cutting. Cutting is generally accomplished in either of two ways. A sidestep cut is one in which the outgoing path proceeds away from the support leg side, whilst a crossover cut results in outgoing motion toward the support leg side. Schot et al. (1995) investigated GRF in 45° and 90° cutting manoeuvres in twelve subjects. Trials took place indoors with subjects wearing standard court shoes directly onto the force platform surface. Vertical and horizontal ground reaction force measures were calculated in addition to the free moment. Significant differences were found between the two severities of cut with the second maximum of the vertical GRF and the average vertical GRF being greater in the 45° movement, whilst the average braking and propulsion forces were greater at 90°. No running velocity constraints were imposed, therefore the increase in vertical ground reaction force (VGRF) at 45° could indicate a higher approach speed. Free moment values exhibited a large coefficient of variation (69%), suggesting different movement strategies may be executed to complete the cutting task successfully (Tibone et al., 1986). The suggestion of torque being the mechanism by which to accomplish cutting manoeuvres (Andrews et al., 1977, cited by Schot et al., 1995) was contested by Hamill et al. (1987), who claimed the change of direction was predominantly created by increased mediolateral force. Schot et al. (1995) concluded that both mechanisms may have been used, but that the increase in mediolateral force was the principal mechanism.

An analysis of mediolateral forces in conjunction with free moment values from ground reaction force measures may resolve this conflict in the literature.

Another form of non-linear motion applicable to soccer performance is that of curvilinear running. The effect of running a track turn was investigated by Hamill et al. (1987) who measured GRF in five male subjects running at 6.31 m/s. Data were collected for outside and inside leg footfalls at a curve radius of 31.5m in addition to straight trials. Results showed significant differences on all vertical variables describing the impact phase, with the outside leg always displaying greater values. However, time to the first maximum was shorter for the inside leg, as was time and impulse to the first minimum. No anterior - posterior differences were observed between the conditions. Differences were noted in all mediolateral forces, with forces in curved running always greater. Force excursions over the support period ranged from 1.206 Ns in the straight condition, to 2.215 Ns for the outside leg. However, over the total footfall, mean excursions were greater for the inside than the outside leg. The data presented provided data concerning curvilinear motion, yet did not cover the range of curvilinear motions found in soccer.

One of the inherent problems in attaining reliable and valid ground reaction force data in soccer is the footwear and surface generally used. The majority of research has investigated ground reaction force relating to soccer movements using either indoor shoes on an indoor runway (Schot et al., 1995) or a force platform covered in artificial turf (Rodano et al., 1988). Whilst some studies have used mechanical equipment to quantify torques on natural grass (Bonstingl et al., 1975), only one study from the literature available has performed subject tests using soccer players in soccer footwear on a natural turf surface. Saggini and Vecchiet (1994) used a force platform covered with natural turf to measure ground reaction force during linear jogging at 2.8 m/s. Although differences between male and female players' ground reaction force were presented, the method of attachment of the surface to the platform, thickness of turf, and footwear used were not detailed. Unfortunately, a description of the force time curves was not presented, so comparison with other studies using soccer players (Saggini et al., 1992c) or those using runners (Cavanagh and Lafortune, 1980; Bates et al., 1983; Clarke et al., 1983; Hamill et al., 1983) was not possible. Once a set of reference data has been substantiated in such conditions, it would appear logical to assess the stress endured at the shoe-surface interface during non-linear and soccer specific movements, ensuring increased ecological validity by the use of standard soccer footwear and surfaces.

It is still the case, however, that the primary use of GRF measurements in soccer have occurred during the assessment of the kicking action. As mentioned in previous sections, the portion of the game where a player kicks a ball is relatively small. Therefore measurement of forces during soccer specific situations must be increased if the demands of the game are to be understood further. The distribution of forces during soccer specific actions could provide valuable data for technique analysis and for footwear development. Also, the proposed mechanisms for curvilinear progression suggested in the literature show a conflict between torque generation at the shoe-surface interface, and increased mediolateral force, which must be resolved. A dominance of straight treadmill and overground running GRF measurements appears in the literature, which needs to be redressed if we are to gain further insight into non-linear movements in soccer and other running based sports.

2.7

SUMMARY OF LITERATURE AND RESEARCH RATIONALE

Biomechanical studies in soccer have focussed on kicking as an important and fundamental skill of the game. Jogging and running are also fundamental to soccer play, yet often these actions occur in a curved path as they are performed to execute tactical manoeuvres and gain a shooting chance at the opponents goal. The cyclical, symmetrical nature of linear running gait has been established by the measurement of muscular activity, stride kinematics and forces generated within each stride (section 2.3). Although many studies were conducted on linear running, there were only a few studies concerning curvilinear motion, and these were related to athletics. The mechanisms involved in the performance of curvilinear motion have therefore yet to be established. Soccer, and other field games also involve other non-linear movements such as cutting and turning to gain tactical advantage, yet little research has focussed on the biomechanics of curvilinear or non-linear motion.

Studies reviewed in electromyography were concerned with linear motion (Elliot and Blanksby, 1976; Nilsson et al., 1985). The changing movement pattern of the body during curvilinear motion is thought to emanate from altered muscular activity at the lower extremity, and posed an important research question. The use of electromyographic techniques to assess the altered muscular activity during curvilinear motion would therefore provide important baseline data in the aim to establish mechanisms of curvilinear motion specific to soccer performance. In linear motion, the majority of the lower extremity muscles that were investigated to date primarily cause movement in the sagittal plane. A concept identified in this thesis (chapter 3) was the adaptive muscular activity that enables curvilinear motion involves muscles which control movement in the frontal or transverse planes of the lower extremity, particularly the hip and ankle joints. The logic for the selection of electromyographic data for the first study in this thesis was that the technique was suspected to also provide an insight into the effects of the specialist shoe-surface interface used in soccer.

Curvilinear motion is also likely to generate altered stride kinematics, though conflicting reports were evident in the studies reviewed in the literature (section 2.4). The concept that a mechanism of curvilinear motion derives from differing stride kinematics for the inside and outside leg is investigated within the research detailed in chapters 3, 4 and 5 of this thesis.

During the performance of linear running overground, measures of ground reaction force have been made during the stance phase and the effects of shoe design and sole configuration investigated by a number of authors (section 2.6). Experimental literature has identified the importance of the medial and lateral force generated in the maintenance of curvilinear motion in athletics. Conflicts from the literature suggest it may be possible that either the muscular forces or moments generated are key to the achievement of motion along a curved path. This conflict should be addressed. In addition the relative contribution of each limb may result in differences in the ground reaction forces occurring at the inside and outside leg of the curve. Such concepts were investigated and subsequently detailed in chapter 6 of this thesis.

The importance of non-linear motion in soccer and the importance of cutting and turning movements in tactical play were considered, with existing literature in kicking also reviewed in section 2.6.1. In soccer play on natural turf, boots with studded outsoles are worn to aid performance. The concept of whether the shoe-surface interface of the soccer boot enabled different application of the forces occurring during ground contact posed an interesting question. To gain further information about the effect of the soccer boot during curvilinear motion, experimental work composing this thesis was performed with subjects wearing training shoes as well as boots. Further investigation of the effect of different soccer shoe sole configuration at the shoe-surface interface would enable insight to be gained into the role of frictional forces in non-linear motion and is reported in chapter 7.

Each study performed during this thesis, which sought to establish the mechanisms of non-linear motion specific to soccer performance, informed and raised questions which were further investigated wherever feasible within the restriction of rational experimental design and available resources.

CHAPTER 3

CHAPTER 3

Lower Extremity Muscle Activity during Linear and Curvilinear Gait While Wearing Training Shoes and Soccer Boots

3.1
INTRODUCTION TO METHODS

This study was undertaken to establish whether altered lower limb muscular activity was a mechanism which enabled curvilinear motion to be performed. Linear motion was performed, in addition to curvilinear motion while jogging and running along paths of different curvature. For ecological relevance to soccer, experimental subjects wore soccer boots and performed on a natural turf surface. In addition, as there was little scientific information on the effect of friction at the boot-turf interface, all experimental trials were repeated in training shoes. This experimental design allowed the prime interest of investigating muscular activity during curvilinear motion. Also, the design allowed the suspected adaptation effect of the soccer boot upon the friction at the shoe-surface interface to be identified.

As straight locomotion is required in the performance of many sports, the application of results from controlled laboratory investigations into EMG activity during gait may be attempted. However, all field games require movements of the athlete that do not fit the typical straight sprinting or running gait patterns described in the literature, and athlete movement will depend upon many factors present in the sporting environment. For example in the game of soccer, the athlete may perform a straight sprint for the ball, cut and weave to pass an opponent, jump for a header and make a curved run in order not to be caught offside, clearly involving locomotion in some form, but not that which could be categorised as walking, running or sprinting as described in previous research. Players may experience the need to change their direction of travel over a period of time, as in bend running in athletics, or almost instantaneously as in side step during field invasion games. Such directional changes occur in all running based sports with the exception of 60/100m track sprinting. Although such movements are commonplace, the description and quantification of the mechanisms employed, which differ to straight motion, have received little attention in the scientific literature (Greene and McMahon, 1979; Stoner and Ben-Siri 1979; Greene, 1985; Hamill et al., 1987), with no data reported of muscular activity.

According to Newton's first law of linear motion, a body continues in its state of rest or motion in a straight line unless compelled to change that state by external forces exerted upon it. During human movement the forces acting on the skeletal body segments are primarily gravitational and internal muscle forces. If these forces were applied in the plane of travel of a body, then segmental accelerations would occur in the same plane. Any forces applied tangentially to this plane would cause accelerations of the body in planes other than that of the direction of travel. Hence, forces applied tangentially to produce changes in direction during the gait cycle may involve muscle activity that is inherently different from that previously reported for linear motion.

In linear motion, the majority of the lower extremity muscles that have been investigated to date, primarily cause movement within the saggital plane and possess very small moment arms in terms of movements outside of the that plane. With the activity of large groups of superficial muscles having previously been described and agreed by a number of authors for straight running (e.g. Mann and Hagy, 1980; Schwab et al., 1983) it is possible that the differences in muscle activity for non-linear, or curvilinear movements would occur in the activity of muscles that control movement in the frontal or transverse planes of the lower extremity. The individual muscles involved in such adaptation needed to be established through pilot testing. The problem required a theoretical approach from the mechanical principles derived from the analysis of figure 1.1.

In the lower extremity, movement can occur at the hip, knee or the ankle joint complexes. However, movement at the knee joint only occurs in the single plane, with flexion and extension the only actions. Therefore, apart from the stabilising functions of the vastus medialis and lateralis during eccentric force production at the knee (Buchanan et al., 1996), muscle activity would remain the same throughout the range of motion of the knee joint. The only exception would be altered activity due to the length change of the rectus femoris and biceps femoris muscles as the femur orientation is altered at the hip joint. Observation of slow motion video recording by the author of this thesis suggested greater variations of segment orientation and limb placement at the ankle and hip joints as an athlete proceeded through a curvilinear gait cycle compared to straight running. Movement appeared to occur outside the sagittal plane to prepare the limb for footstrike, and place the body at an optimum orientation to maintain curvilinear motion. In order to describe accurately non-linear motion in soccer, the present investigation focused on the actions occurring in curvilinear motion, which result in movements of the lower extremity outside of the sagittal plane. Therefore, in an attempt to identify the major mechanisms enabling the performance of curvilinear gait, the muscles chosen for investigation were located at the hip and ankle joint complexes. The final muscles selected for the main experimental study (hypothesis 2) were established during extensive pilot testing.

Variations in muscle activity would be monitored under different conditions of curve, athlete velocity, and shoe-surface frictional qualities. To determine the muscle actions occurring during the curvilinear gait cycle, this research aimed to provide quantitative information concerning the onset and cessation of the targeted muscle activity. Such measures were considered the prime approach to the investigation, as patterns of activity in lower extremity muscles have been established as cyclical in nature in straight running eg. Elliot and Blanksby (1979). Therefore, it was anticipated that the disproportionate distances of travel for each limb in the curved gait cycle would alter subtle timing of muscle actions to maintain the cyclic nature of the activity. Hence, temporal measures of muscular activity would be used as the prime measure for change in muscular activity throughout the study. The expected change in temporal muscle activity as a mechanism of performing curvilinear motion therefore formed one of the experimental hypotheses (hypothesis 1).

Changes in EMG magnitude will indicate only a difference in muscle force, which could be achieved by simply increasing gait velocity, and therefore would not describe the difference in type, or mechanism of curvilinear motion. To maintain motion at the same velocity, yet at a tighter grade of curve, muscular force would be expected to increase. Therefore, an increased magnitude of muscular contraction was hypothesised as adaptation to curvilinear motion (hypothesis 3).

The use of soccer specific footwear enables penetration of the studs into the natural turf surface. When performing curvilinear motion one may expect that the medial side of the outside boot, and the lateral side of the inside boot would penetrate the turf to a greater extent than the corresponding sides. If such a suggestion were true, one would expect the position of the ankle to be less pronated or supinated in soccer boots than when using training shoes. From this notion one can thus hypothesise that the amount of muscular activity around the stance phase will decrease when wearing soccer boots as opposed to training shoes during curvilinear motion (hypothesis 4).

Hypotheses

1)

H_1: Temporal muscle activity in those muscles monitored will exhibit adaptation to curvilinear running when compared to linear running.

2)

H_2: Temporal muscle activity in the gluteus maximus, tensor fascia latae, gastrocnemius, peroneus brevis, and tibialis anterior will show statistically significant adaptation as the grade of curve becomes more severe.

3)

H_3: The magnitude of muscular contraction as measured by EMG amplitude will increase as adaptation from linear to increased severity of curvilinear motion occurs.

4)

H_4: The amount of muscular activity around the stance phase will decrease when wearing soccer boots as opposed to training shoes during curvilinear motion.

3.2
METHODS

3.2.1 Pilot work

Pilot work was required to finalise the experimental protocol for the main study. As no previous research was available in this area, investigation into movement patterns, instrumentation, muscle selection and procedure needed to be undertaken before the main study took place.

Movement Patterns

As many soccer movements are very rarely identical it was necessary first to establish muscle activity of curvilinear motion under controlled experimental conditions. Information was available regarding distances covered in a competitive soccer match (Reilly and Thomas, 1976; Yamanaka et al., 1988), with a division of time and distance encompassed by different intensities of movement (walk, jog, cruise, sprint, backwards, sideways). However, the discrete categorisation of movement did not provide information concerning the nature or frequency of non-linear actions. Knowledge of the type of activity performed in a competitive soccer game was necessary before experimental design could take place. Observation of video and live soccer matches provided information of the variation in curvilinear motion that occurred during competitive situations. Instantly notable non-linear motion occurs in the form of jumping, tackling, jockeying and turning with the ball. Such activities are situation dependent in terms of the speed and distance covered in their performance, making their standardisation complicated, and therefore did not present simple avenues of scientific investigation. Movements of a curvilinear nature, such as a run to occupy space, or a run to maintain speed but remain onside, provide typical motions that can be more easily standardised under experimental conditions. It was estimated from observation of video tape that the curved runs made by players increased in radius from a minimum of approximately 5 meters. Curvilinear runs were observed at differing grades, with the slightest curve being estimated at a radius of approximately 15m.

It was realised that a primary factor in the performance of non-linear actions was the friction at the shoe surface interface (Ekblom, 1993) which also influenced the degree of movement of the centre of gravity. It is possible that if friction coefficients altered non-linear performance, different footwear could similarly alter movement, and hence muscle activity, under the same conditions. Such a suggestion forms the basis of the final experimental hypothesis. The sport of soccer offers a choice of different footwear dependent on surface conditions, with studded footwear generally being used in wet conditions, whilst flat soled soccer shoes with a contoured sole would be used when ground conditions were too hard to enable penetration of a studded or moulded boot. Comparison of such extremes of studded and flat sole design therefore presented a method of investigation of frictional variation.

The speed of movement of a player will be affected by many factors in a game situation. Whether a player is directly or indirectly involved with play, has limited amount of space available in which to move, or is in or out of position will all effect movement speed. To separate movement into speed categories, subjective responses of soccer coaches were obtained from observation of an English Premier League game recorded on videotape. The coaches (n=2) classified players' movement into four categories; walk, jog, run, and sprint.

All variations in movement warranted investigation, but differing velocities must also be used to gain a global picture of muscle activity patterns. It was decided to use two gait velocities for the study. The two velocities of jogging and running were selected as it was evident that the most curved motion, 5m radius, occured only whilst players were jogging or running in competitive situations. To perform such a tight movement whilst sprinting did not appear possible. Sprinting accounted for linear motion with little if any rapid angular change, as shown in 200m athletics, where sprint performance entails sprinting the inside lane of the bend of approximately 17m radius. An approximation of two velocities of motion were then made by the timing of three soccer players through light gates placed 3m apart. For each trial the player was given a verbal instruction of 'jog' or 'run' from a position 10m in front of the light gates.

Five trials per subject were completed. Self-selected means were 4.40 m/s and 5.40 m/s for jogging and running respectively.

Six muscles (gluteus maximus, tensor fascia latae, gastrocnemius, peroneus brevis, and tibialis anterior) were selected for monitoring on the basis of results from pilot testing. These muscles were monitored only on the right leg of each subject. Hence, to obtain information on both the inside and outside legs during curvilinear motion, it would be necessary for the players to complete two trials at each curved radius. Clockwise and anticlockwise travel along curvilinear paths would then enable data capture for both inside and outside limbs.

Environment

To maximise the ecological validity of testing conditions, the experiment was conducted outdoors on a flat, natural turf surface in an area of approximately 30 x 20m, with grass length regularly cut to ensure replication of soccer pitch conditions. All testing was completed in dry atmospheric conditions due to the need to use electrical equipment outdoors. The surface was irrigated to enable penetration of a standard soccer boot stud design (six-stud). Irrigation commenced approximately 3-4 hours before the collection of data to ensure sufficient turf moisture. Irrigation was monitored constantly with alteration of irrigation position occurring approximately every twenty minutes. Turf moisture was assessed with a soil wetness meter (Rapitest, UK) with experimental trials proceeding only if a minimal value of 3 was obtained (range 1-4). The basis for the nominated value of the wetness readings was achieved from a subjective assessment by three college soccer players prior to the study.

Instrumentation

Limitations of the electromyography equipment meant that eight channels were available for the recording of data. For comparison of results to those of Elliot and Blanksby (1979), and Mann and Hagy (1980), reference points were required to indicate the start and end of the gait cycle. Heel strike was chosen to indicate the beginning of the gait cycle, with footswitches located within cut-out sections of the insole. With two channels used for input of binary footswitch data, six channels were available for the recording of muscle activity. Medicotest N-50-E electrodes were applied to six muscles of the right leg. Surface electrode pilot work on sampling frequency led to EMG data sampled at a frequency of 500Hz using a radio telemetry system (MIE MTR8; Leeds, England) with a Yagi aerial. Data were recorded on a Viglen 4DX33 personal computer running orthodata GmbH MYO-DAT 3.0 software for MIE MT8-MBM. The investigation focused on the onset and cessation times of muscle activity within the gait cycle. Therefore, as data for muscles under each experimental condition would be presented as a percentage time in the gait cycle, results could be compared within and between subjects, as the variation of stride length and cadence were negated.

EMG Electrode Application

For preparation, the skin was first shaved over the belly of the target muscles to ensure good electrode attachment. Then skin was cleaned with an alcohol swab, and rasped with a single subject disposable Medicotest skin rasp made of a Velcro line material, before swabbing again to remove dead skin cells. An electrode gel was then placed on the skin area and left for 3-4 minutes before being removed. The skin area was then re-swabbed and dried before the disposable electrodes were attached to the muscle belly along the line of the muscle fibres. Electrode spacing was 4cm for the peroneus brevis due to the smaller superficial area, and 5cm for all other muscles. The reference electrodes for these muscles were positioned on the patella and lateral malleolus of the ankle on the right leg. Individual muscle amplifier boxes were taped tightly to the skin surface to minimise unwanted movement artefact. Skin impedance was assessed prior to testing and electrodes re-applied if a reading exceeded 10 kΩ. Individual muscle amplifier box leads were connected to the telemetry transmitter unit worn around the subject's waist on a woven belt. Maximum voluntary contractions were performed for each muscle by the subject before and after the testing period according to procedures outlined by Daniels et al. (1956) to enable confirmation of similar levels of muscular force at each grade of curve and validate the use of temporal activity values as the prime measure.

Muscle Selection

The muscles selected for investigation received continuous modification, as feedback was gained from feasibility trials. As the focus of the investigation was an examination of the differences from linear gait, some muscles were targeted that would be involved in movement outside of the saggital plane. The muscles considered in the study were:-

Gluteus Maximus (GLUT)
Tensor Fascia Latae (TFL)
Rectus Femoris (RF)
Biceps Femoris (BF)
Tibialis Anterior (TA)
Gastrocnemius Medial (GM) and Lateral (GL) heads
Peroneus Longus (PL)
Peroneus Brevis (PB)
Flexor Digitorum Longus (FDL)

Procedure

One soccer player (age 23 yrs) was studied during pilot and feasibility work. The subject consented to participate (Appendix G) and the subject wore standard six-stud soccer footwear. Activity from the muscles considered was monitored during conditions of straight and curvilinear motion. Linear speed of the subject was measured using infra-red light gates (Cla-Win Timer, University College Chichester). The subject completed bouts of activity at the pre-determined linear velocities equating to jogging and running around curvilinear paths marked with cones at radii of 5, 10 and 15m. Data were then sampled for five second periods in each trial, encompassing the passage of the subject through the light gates. The series of trials were then repeated at each velocity and radius in the opposite direction to enable acquisition of data relating to both the inside and outside leg during curvilinear motion.

Results and Discussion

Feasibility investigations attempted to target muscles used by previous authors such as Elliot and Blanksby (1979) who monitored rectus femoris and biceps femoris muscles. However, no noticeable difference in onset and cessation was evident between conditions, therefore it was decided for the present study that target muscles ought to display functions that cause movement outside of the saggital plane. At the hip joint, internal and external rotation were of interest therefore the tensor fascia latae and gluteus maximus were selected due to their function, superficial location and relatively large size. As the tensor fascia latae also causes hip flexion, it was the posteriolateral fibres of the muscle that were targeted (Pare et al., 1981).

The muscles investigated in the pilot study and the criteria involved in the selection of final muscles for the main study are listed in Table 3.1.

MUSCLE	RESULT
Selected	
Tensor Fascia Latae	Activation during abduction of the leg
Gluteus Maximus	Differences with extension (speed) and abduction (grade)
Tibialis Anterior	Indicator of inversion, dorsi-flexion.
Peroneus Brevis	Indicator of foot eversion. Differences with grade of curve.
Gastrocnemius -lateral -medial	Temporal firing pattern altered comparing inside to outside leg
Rejected	
Rectus Femoris	No difference in temporal activity during curvilinear motion
Biceps Femoris	No difference in temporal activity during curvilinear motion
Flexor Digitorum Longus	Inversion of foot. Only active in stabalising ankle at stance.
Peroneus Longus	Extraneous activity from localised muscles. Location difficult.

Table 3.1 Results of investigation to determine muscles to be used in main study

Feasibility trials also indicated that at the ankle joint movements of plantar and dorsi flexion, inversion and eversion were of relevance. Tibialis anterior was targeted for functions of inversion, dorsi flexion, and accessibility of monitoring. The flexor digitorum longus was investigated due to its involvement as a prime mover in inversion. However, it proved only to be active for the duration of foot contact in a stabilising mode, showing no modification of activity under different curvilinear conditions.

For indication of ankle eversion the peroneus longus was targeted as the designated muscle. Feasibility work once again showed this to be inappropriate as interference from localised muscles disrupted the signal. The electrodes were repositioned at various locations along the peroneus longus muscle before it was realised that an alternative was required. So, the peroneus brevis was chosen in preference to the peroneus longus for ease of location of the muscle belly, and also reduced interference from other localised superficial muscles. Tomaro and Burdett (1993) found extraneous noise from gastrocnemius and tibialis anterior whilst trying to monitor peroneus activity in the brevis muscle, whereas the present investigation monitored brevis activity successfully but found additional extraneous muscle activity in the peroneus longus muscle.

The remaining two channels monitored the gastrocnemius muscle on its medial and lateral heads to assess stabilising functions during ankle inversion/eversion. The notion that these two heads of the muscle could differ in activity came from a feasibility testing observation, indicating that during performance of the more extreme grades of curve the heel of the foot sometimes does not contact the ground and the foot was planted at an angle to the direction of travel. Pilot trials showed slightly earlier activation of one head of the muscle during each direction of travel, justifying the inclusion of the muscle in the main study.

Temporal data values from two consecutive trials are presented below to demonstrate the reliability of these values. Values presented are initial onset times and final offset times of these muscles presented as a percentage of the stride cycle.

	Straight jog 1	Straight jog 2	Straight run 1	Straight run 2
TA on	17.4	21.1	17.7	18.8
TA off	75.4	73.2	82.3	81.3
PB on	24.6	25.4	14.5	14.5
PB off	78.3	78.9	80.6	80.6
GL on	21.7	23.9	19.4	20.3
GL off	89.9	95.8	79	81.3
GM on	21.7	23.9	21	20.3
GM off	79.7	78.9	78.8	71.9
TFL on	17.4	16.9	11.3	12.5
TFL off	91.3	88.7	90.3	92.2
GLUT on	17.4	16.9	11.3	12.5
GLUT off	84.1	87.3	88.7	81.3

Table 3.2 Electromyographical onset and offset values of a single subject (as a percentage of stride).

Table 3.2 shows that for both jogging and running trials show temporal values to be repeatable on the day of testing. The only exception was the offset times for the lateral head of the gastrocnemius which were slightly different between trials. However, the general pattern showed good replication of temporal values.

3.22 Main Study

Subjects

Ten male college soccer players were selected for the study (mean age 23.7 ± 4.14 years), based on their familiarity with the curved running patterns experienced in the sport. No subjects reported suffering musculoskeletal injuries and all were in good health at the time of testing. The experimental procedures were explained to the subjects and written consent to participate was obtained. The study had received ethical clearance from University College Chichester. All subjects were reminded that they had the right to withdraw from the experimental study at any time. All subjects wore standard six-stud soccer boots and soccer style training shoes. The testing conditions remained the same as adopted in the pilot study.

Instrumentation

For EMG data collection, Medicotest N-50-E electrodes were attached to the prepared skin overlying the belly of tibialis anterior, peroneus brevis, lateral and medial heads of the gastrocnemius, gluteus maximus and tensor fascia latae muscles of the right leg. The actions of these muscles are summarised in Table 3.2.

Muscle	Action (Primary, Secondary, Tertiary)
Gluteus Maximus	Extension of femur; abduction of femur; lateral rotation of femur
Tensor Fascia Latae	Flexion and abduction of femur; medial rotation of femur
Gastrocnemius	Plantar flexion of foot; flexion of knee
Peroneus Brevis	Plantar flexion of foot; eversion of foot
Tibialis Anterior	Dorsi flexion of foot; inversion of foot

Table 3.3 Action of targeted muscles

Binary footswitches were positioned under the heel on the plantar surface of the foot, to assess the instant of foot contact with the ground and enable stride data to be normalised with respect to stride cycle timing.

An infra-red timing system (Cla-Win timer, University College Chichester) was used to quantify the linear velocity of the subject in each trial. Infra-red timing gates were positioned 3m apart at hip height to calculate linear velocity on the runway. Video recording of the subject during each trial took place using two video recorders (Panasonic VHS Supercam AG-DP800E, EG) positioned to view the gait motion for reference. One camera was positioned to film the rear view of the frontal plane as the subject passed through the timing light gates on the curvilinear path. A second camera provided a visual record of the whole experimental procedure and an audio record of the experimenter's comments.

Procedure

Experimental procedure remained as described in the pilot study. Subjects performed trials at a jogging velocity of 4.4 m/s and a running velocity of 5.4 m/s (+/- 5%). Subjects repeated the series of trials in the opposite direction, therefore collecting data for the inside and outside limbs. Camera positions were also reversed for these conditions. Trials were performed around three circular sections of 5, 10, 15 metre radii circles marked on the natural grass surface with cones. Six muscles were monitored and two binary footswitches used to identify heel strike. Jogging and running trials at each of the three radii in both directions gave a series of 12 trials for each subject. Following the completion of these trials each subject performed 2 straight trials at the predetermined velocities for comparative purposes. The series of 14 trials were then repeated for the alternative footwear condition, with the order of footwear randomised.

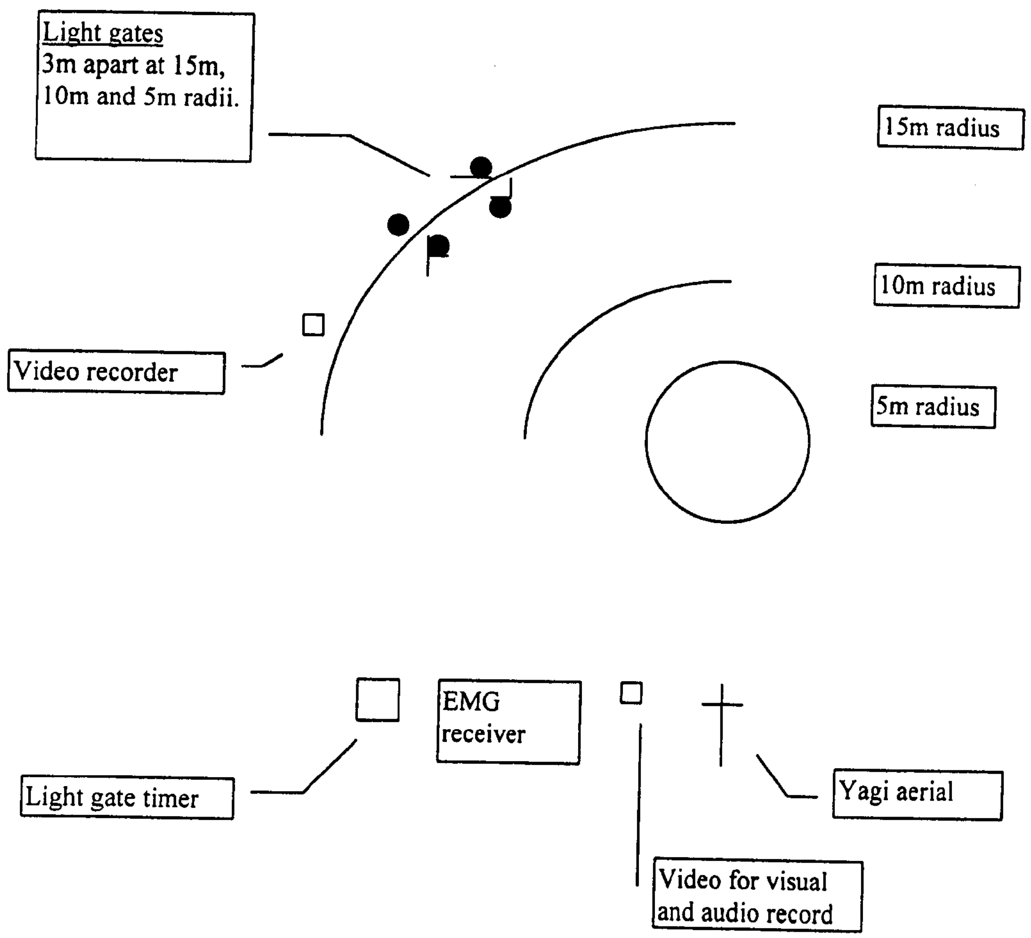

Figure 3.1 Experimental set-up

Data Analysis

Data acquisition occurred over a five second time period encompassing the subjects motion through the light gates. EMG activity was recorded for approximately ten strides to enable analysis of a typical stride in every data set. A typical stride in each trial was identified as showing low noise levels and was located approximately as the subject passed through the light gates.

Onset and cessation times were recorded of each burst of EMG activity throughout the stride duration by cursor movement in the Orthodata software, with activity onset and cessation identified manually. Times were noted for each muscle over a one stride period (right heel strike to right heel strike); pulse durations were calculated also. Each time value was normalised to percentage of stride to enable inter and intra individual comparison. For each condition one stride was analysed, with the exception of the straight motion where two strides were used and a mean value taken. Two strides were averaged in the straight condition as these trials were to serve as the baseline data set. Data were entered into spreadsheet format (Windows Excel 5.0) and temporal variables of burst onset, cessation and duration averaged for the ten subjects. For the purpose of these investigations, the term 'grade' was used interchangeably with the word 'radius.' A three factor (grade, leg, speed) analysis of variance was performed within each subject to determine any significant differences. Data sets for soccer boots and training shoes were analysed separately, and differences then compared between each.

The magnitude of muscle activity was considered of less importance in the adaptive mechanism because an alteration in the magnitude would not suggest a change in performance mechanism, simply a change in applied muscular force. Maximal enveloped EMG values were measured for each muscle during each straight and 5m trial, with a two factor (speed, leg) analysis of variance performed within each subject to determine significant differences.

It was clear from initial scanning of raw data that 85% of data for all muscle groups had identifiable recruitment patterns. Statistical analysis was performed on the 85% of data, outliers to which will be described later in this chapter. The adaptations in curvilinear motion with respect to straight running motion in the EMG temporal and magnitude activity are considered using the following comparisons:-

Straight vs. curvilinear motion in shoes (section 1)

Straight vs. curvilinear motion in boots (section 2)

Differences of shoes to boots (section 3)

3.3
RESULTS

3.3.1

COMPARISON OF STRAIGHT AND CURVILINEAR MOTION IN SHOES

All of the literature related to electromyography in running from a sports biomechanics perspective has concerned linear motion in soccer style training shoes. The results in the following section show how data collected for linear motion compares with values for differing grades of curvilinear motion. These data will also be contrasted with values from the literature on linear treadmill and overground running.

In order to condense and present information in a concise and comprehensible format, the average muscle recruitment pattern for each trial condition was represented as a graphical bar output, calculated as percentages of the stride cycle. These data show the mean temporal data for a test condition. Right foot heel strike occurs at 0% and 100% of stride.

Figure 3.2 indicates that the initial muscle activation pattern during straight running showed all muscles exhibited activity around the stance phase, with cessation of activity in the order of TFL, GLUT, TA, PB and gastrocnemius. TFL and TA also displayed a mid cycle swing phase pulse. TFL activity could represent the anterior fibres of the muscle firing to flex the thigh during the swing phase whilst TA swing phase activity could correspond to dorsi flexion in preparation for heel strike.

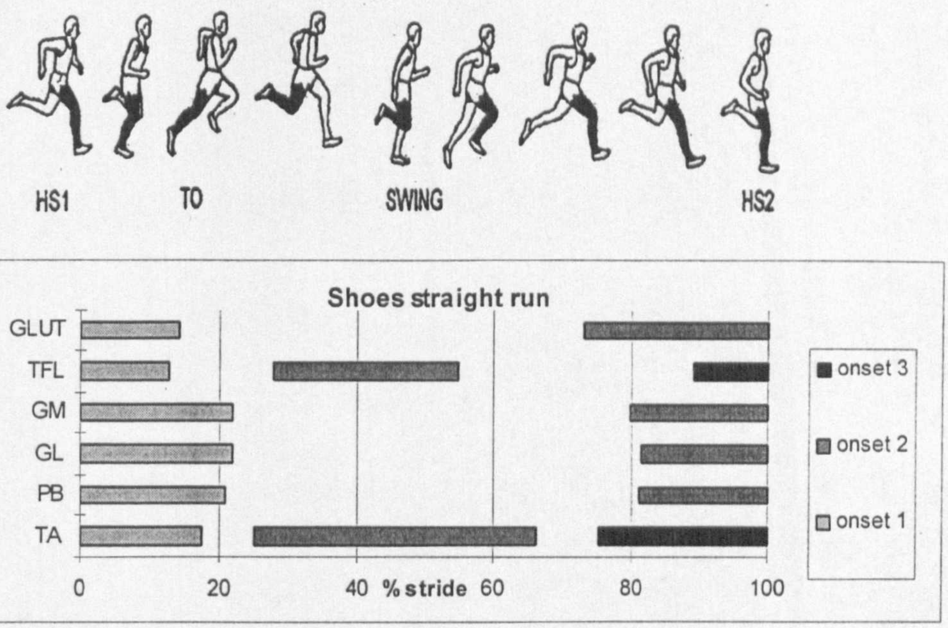

Figure 3.2 Mean temporal EMG activity for straight running in flat soled shoes (10 subjects).

NB. HS1 corresponds to 0% stride and HS2 to 100% stride respectively.

Possibly all selected muscles were being used to stabilise the joints of the lower extremity during the stance phase. With the exception of the TFL, these muscles would also provide propulsive force for the subsequent stride. Activation prior to heel strike occurred progressively from approx 68% cycle time in GLUT, TA, GM, PB, GL and TFL.

For the straight run it was assumed that these activity patterns were representative of both left and right legs, and therefore will be used in comparing inside and outside leg activity in curvilinear motion. It should be noted that graphical displays of mean temporal values show a complete stride cycle from right heel strike to right heel strike. Therefore, a muscle showing activity at 100% of the stride cycle would continue at 0% in the next cycle. This period of activity about heel strike was modified at different grades of curve, and will be referred to as duration at stance (DUS).

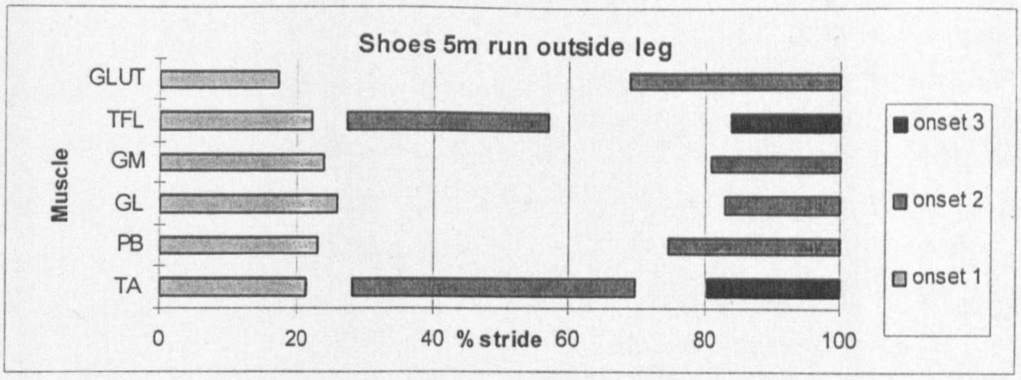

Figure 3.3 Mean temporal EMG activity for the outside leg running in shoes at 5m radius (10 subjects).

Comparison of the temporal activity in the outside leg at 5m (Fig. 3.3) to the straight running condition in figure 3.2, indicated that overall activity had increased. The increase occurred predominantly around the stance phase, where duration at stance had increased in each muscle. Such increased temporal activity in the outside leg was considered a function of curvilinear motion. These data, combined with further information on foot contact times, will provide an insight into the mechanisms of curvilinear motion.

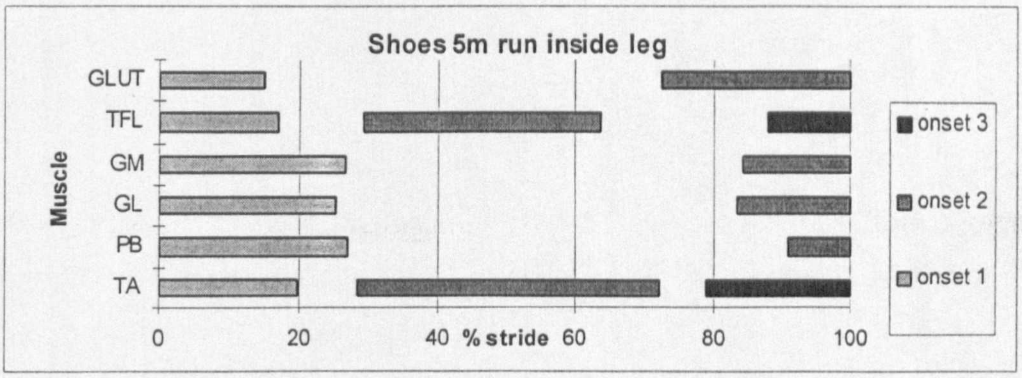

Figure 3.4 Mean temporal values for the inside leg running in shoes at 5m radius (10 subjects).

Figure 3.4 indicates that overall greater activity was shown in the inside leg compared to the straight condition around the stance phase. Most notable was an increase in the duration of the swing phase pulse in the TFL and TA muscles. Such findings would suggest increased swing phase activity to be an adaptive response to curvilinear motion in the inside leg.

A greater muscular adaptation was noted at the 5m radius than other grades of curve. In addition it appeared that an overall greater adaptation had occurred in the outside leg during curvilinear motion. Such adaptations were based around the stance phase in the outside leg, whereas adaptation in the inside leg was predominantly noticeable in the swing phase.

To determine whether the muscular adaptations observed were due to alterations in stride pattern as well as direction of travel, selected kinematic variables were calculated. Kinematic measures of stride length and stride frequency (cadence) were derived from subject velocity and period of stride. Velocity was measured from the infra-red timing system whilst period of stride was taken from the mean of stride cycles in each five second trial.

$$\text{Velocity} = \text{Stride Length} \times \text{Stride Frequency}$$

Average stride length and cadence values were calculated from approximately 5-10 strides, dependent on the quality of data, using footswitch activation and associated velocity data for both jogging and running. Alterations in stride kinematics showed player velocity tended to reduce with increasing grade of curvature. As the grade of curve became tighter the stride length was reduced, accompanied by an increase in stride cadence. Such adaptations were most noticeable at the 5m grade of curve. Significant differences ($P < 0.001$) were shown between stride length and cadence with grade of curve. Velocity was also shown to alter significantly ($P < 0.001$) between grades of curve, reducing the significance of the adaptations in stride length and cadence with differing grade of curve. This possibly occurred due to insufficient friction at the shoe-surface interface. Differences in stride length and cadence may then be attributable to altered subject velocity rather than adaptation to differing curvature.

	JOG RADIUS				RUN RADIUS			
	0m	15m	10m	5m	0m	15m	10m	5m
velocity. ms^{-1}	4.73 ±0.018	4.40 ±0.001	4.53 ±0.001	4.31 ±0.001	5.98 ±0.014	5.40 ±0.001	5.40 ±0.001	4.95 ±0.001
frequency. s^{-1}	1.47 ±0.024	1.49 ±0.002	1.50 ±0.002	1.52 ±0.002	1.64 ±0.041	1.64 ±0.003	1.63 ±0.003	1.72 ±0.003
length m	3.14 ±0.076	2.99 ±0.006	3.05 ±0.009	3.02 ±0.011	3.73 ±0.114	3.35 ±0.011	3.34 ±0.008	2.98 ±0.010

Table 3.4 Mean kinematic variables and standard errors at differing curvilinear grades in shoes

Curvilinear trials showed an EMG activity pattern that was inherently different from straight running in shoes. Temporal EMG adaptations were evident at all grades of curve, but were more significant at the tightest 5m radius. Tables 3.4 and 3.5 show significant differences in the data, with levels of significance indicated by

☆ $P \leq 0.001$; ① $P \leq 0.01$; ② $P \leq 0.02$; ③ $P \leq 0.03$; ④ $P \leq 0.04$; ⑤ $P \leq 0.05$;

and DUS representing duration at stance.

	Inside leg muscle						Outside leg muscle					
TEMPORAL RUN EVENT	GLUT	TFL	GM	GL	PB	TA	GLUT	TFL	GM	GL	PB	TA
Offset 1 later at 5m		☆	☆					☆	☆	☆		
Offset 1 later at 5m than straight					①						①	☆
DUS greater at 5m		⑤						⑤				
DUS greater in 5m than straight							⑤					
DUS greater in curved			⑤								⑤	
Onset 2 later at 5m than straight						③						③
Onset final pulse later at 5m												③

Table 3.5 Temporal differences in EMG activity when running on straight (0m), and curvilinear paths of 15m, 10m, 5m radius in flat soled shoes

Table 3.4 shows that the main differences in muscle activity during curvilinear running occurred in the outside leg. The most significant change taking place in the 5m trials was the delayed offset of the first pulse. Many of the changes were evident in both legs e.g. TFL. The outside leg showed increased activity at stance in all muscles when running. TA was the only muscle to show increased activity during swing in running (Table 3.4).

In jogging modifications to the temporal pattern occurred in the outside leg predominantly with no muscles exhibiting identical adaptations in both legs (Table 3.5). Graphical EMG output for mean temporal values in the flat soled shoe condition suggested that the inside leg showed increased activity in swing. However, such observations were not shown to be significant in statistical terms except for TFL ($P<0.01$).

	Inside leg muscle						Outside leg muscle					
TEMPORAL JOG EVENT	GLUT	TFL	GM	GL	PB	TA	GLUT	TFL	GM	GL	PB	TA
Offset 1 later in 5m		☆	☆					☆	☆			☆
Offset 1 later at 5m than straight					①					☆		
DUS greater in 5m		⑤			⑤		⑤	⑤		⑤		
DUS greater in 5m than straight											⑤	
DUS greater in curved						⑤						
Onset 2 earlier in 5m than straight		①						①				
Onset final pulse earlier at 5m								①			④	

Table 3.6 Temporal differences in EMG activity when jogging on straight (0m), and curvilinear paths of 15m, 10m, 5m radius in flat soled shoes.

Similarities between the flat soled shoes data during jogging and running were the later first offset and increased duration at stance (5m) in the GLUT outside leg, GL outside leg, and TFL of both legs. The offset of the first pulse only was increased at 5m for the GM of both legs, PB inside leg, and TA outside leg. PB showed a greater duration at stance in the 5m condition, outside leg, at both velocities.

Major adaptations to curvilinear motion could be considered to be those which were evident in jogging and running as the grade of curve was altered. Several variables displayed such adaptations at the 5m radius:-

Both legs:	GM -	later first offset after heel strike in 5m
Outside leg:	GLUT, TFL, TA	" " " " " " " "
	TFL	DUS greater at 5m
Inside leg:	PB	later first offset after heel strike in 5m

It should be noted that the data presented showed a consistent adaptation to curvilinear motion. Overall, the data show more adaptation taking place in the outside leg of the curve. The outside leg must travel further to reach its next footstrike than the inside leg, a feature of the gait pattern which may cause some of the adaptations observed. The adaptations generally took the form of a later offset of the first pulse of activity following heel strike as an indicator of general shift of activity to later in the cycle.

The swing phase pulse also highlighted consistency in the TA muscle between running and jogging conditions. The onset of the second pulse was later in the 5m condition as subjects were running, but not when jogging. Such a difference could be attributable to the altered stride kinematics of shorter stride length and greater stride frequency as subjects performed the 5m trials.

To express the level of muscular activity during the stride cycle, the maximum amplitude of enveloped EMG values were recorded. If the maximum amplitude of the enveloped signal was not different between grades of curve the use of temporal measures as the prime descriptive variables would be justified. Amplitude data were expected to alter greatest between the straight and 5m radius conditions, therefore only these values were compared. Maximal enveloped EMG values were measured for each muscle during each trial, with a two factor (speed, leg) analysis of variance performed within each subject to determine significant differences.

As expected levels of muscular activity measured by maximum amplitude of enveloped EMG overall were higher when running than jogging (P = 0.028). Significant differences (P < 0.05) were highlighted for the TA swing phase in the outside leg and the straight condition. Similar differences were evident in the swing phase of the TFL muscle. However, no significant differences between straight and curvilinear motion at tightest curvature were found for either the inside or outside leg in running or jogging, suggesting the size of the muscle contraction was not altered by placing the subject on a curvilinear path as opposed to a straight path.

	5m jog inside	5m jog outside	straight jog	5m run inside	5m run outside	straight run
TA st	66.9±1.86	58.9±4.1	70.5±2.6	64.2±3.1	60.8±3.6	73.7±3.8
TA sw	62.3±1.89	60.9±4.0*	54.9±3.4*	63.0±4.7	69.2±4.0*	71.3±3.0*
PB	74.9±11.7	83.3±13.3	109±14.5	71.8±15.6	89.0±12.5	103±10.2
GL	92.6±6.5	97.3±8.5	98.7±8.8	80.6±11.0	95.7±7.8	92.9±5.8
GM	99.2±7.8	93.5±8.4	97.7±12.4	92.1±6.2	88.5±10.2	95.3±10.9
TFL st	32.8±5.9	47.1±8.3	47.1±6.1	32.6±5.9	53.3±7.4	47.5±6.0
TFL sw	47.9±5.8	35.8±4.0*	35.7±5.1*	57.4±5.5	54.8±8.8*	69.5±21*
GLUT	69.7±13.6	77.5±13.8	75.1±12.7	97.4±23.3	99.3±24.9	105±22.7

Table 3.7: Mean ± S.E maximal enveloped EMG expressed as a % MVC for curvilinear motion in shoes. (st = stance; sw = swing, *P<0.05)

In summary, curvilinear motion in flat soled shoes was associated with temporal muscular adaptations in all monitored muscles within a right heel strike to right heel strike stride, except for the GLUT inside leg when straight motion was compared with curvilinear motion at a 5m radius. At the tighter grade of curve there were indications of a reduction in velocity and stride length and an increase in stride frequency (Run:

velocity 17%, stride length 5%, stride frequency 5% Jog: velocity 9%, stride length 3%, stride frequency 3%). Increased levels of muscular contraction amplitude were evident in running compared to jogging. Performance of curvilinear motion did not appear to alter the magnitude of muscular activity in either jogging or running.

Reliability of EMG traces for one subject was presented in section 3.2.1 in the form of onset and offset times of all muscles monitored.

3.3.2

COMPARISON OF STRAIGHT AND CURVILINEAR MOTION IN BOOTS

No previous studies have attempted to capture muscle activity data whilst subjects were wearing soccer boots. Such data would provide important baseline measures when attempting to describe the mechanisms occurring during curvilinear motion in soccer. Results in this section show EMG data from straight motion in soccer boots. These results are also compared with values from curvilinear motion at different grades.

Muscle activity data were captured as raw EMG traces. A representative raw data trace of eight channels can be seen in figure 3.5. It was clear from initial scanning of raw data that 85% of data for all muscle groups had identifiable recruitment patterns. Initially analysis will concentrate on that 85%. Following Kameyama (1990) the remaining 15% of trials (42 across 4 subjects of total 10) are described afterwards.

Data have been presented as a graphical bar output to aid comprehension. These data show the mean temporal data for a test condition with right foot heel strike at 0% and 100% of the stride cycle.

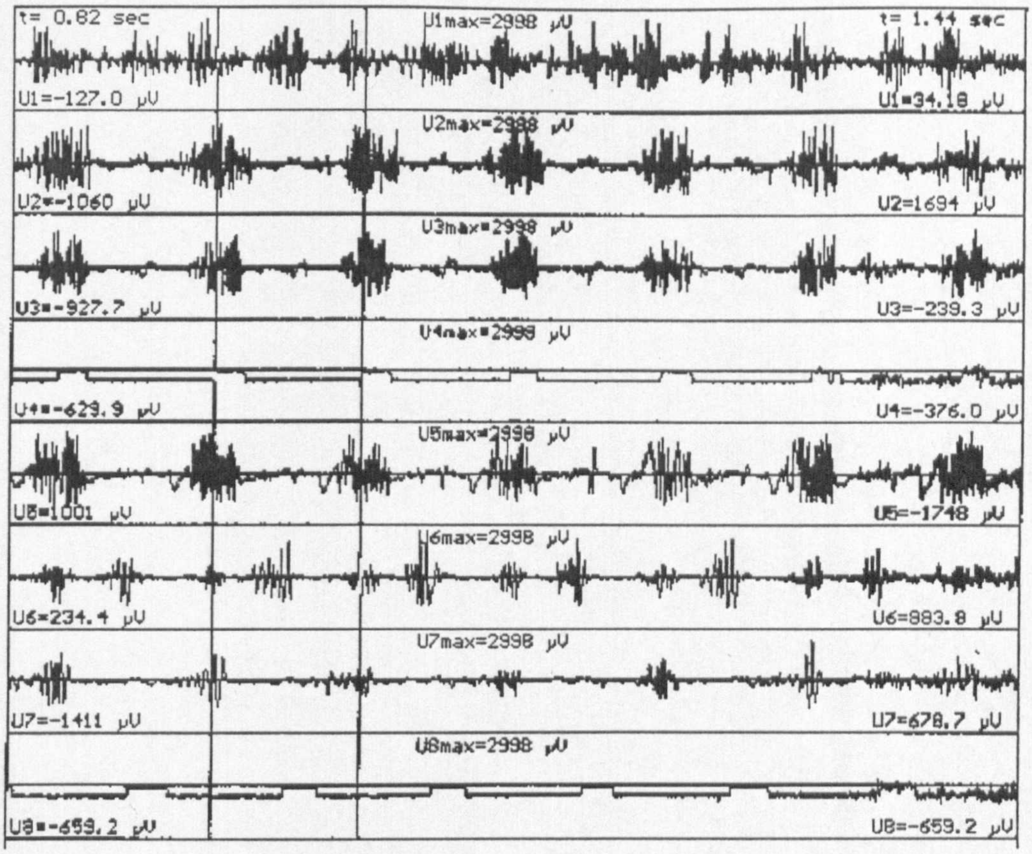

Figure 3.5 Raw data Boots Straight Run

Channel 1: Tibialis Anterior

Channel 2: Peroneus Brevis

Channel 3: Gastrocnemius Lateral Head

Channel 4: Right Footswitch

Channel 5: Gastrocnemius Medial Head

Channel 6: Tensor Fascia Latae

Channel 7: Gluteus Maximus

Channel 8: Left Footswitch

Figure 3.5 shows raw EMG data during a straight running trial in soccer boots. Event markers are placed at a selected stride cycle to indicate heel strike 1 at 0.82 seconds, and heel strike 2 at 1.44 seconds. These data were sampled at 500Hz. Muscle contraction amplitudes are directly comparable as each channel is scaled to 2998μv maximum.

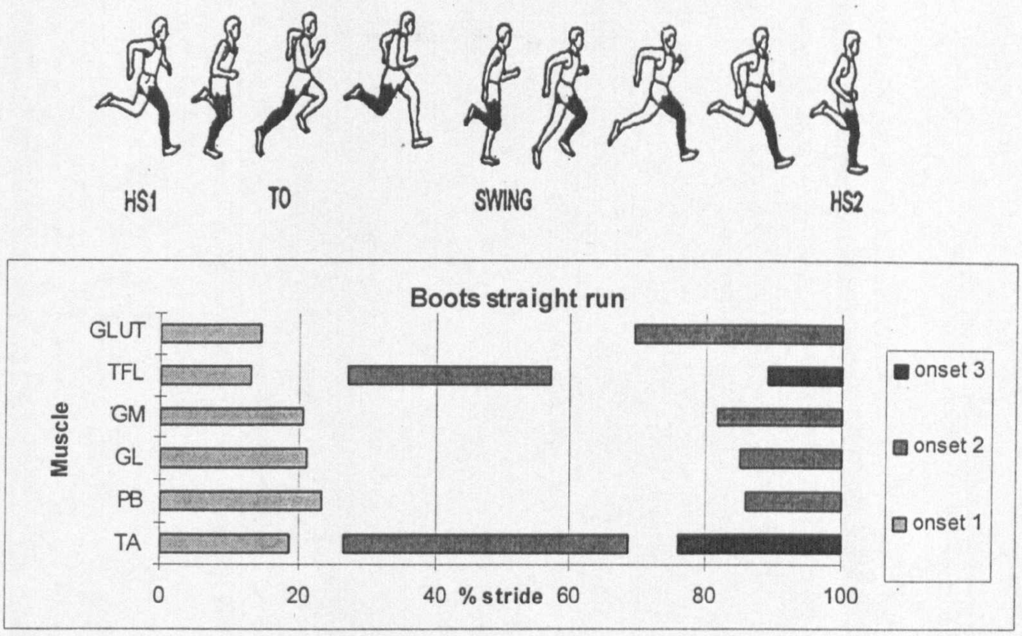

Figure 3.6 Mean temporal EMG activity for straight running in boots (10 subjects)

Figure 3.6 shows the order in which muscle groups ceased firing; Tensor Fascia Latae (TFL), Gluteus Maximus (GLUT), Tibialis Anterior (TA), Gastrocnemius (GM and GL respectively), and finally Peroneus Brevis (PB). Two of the six muscles exhibit a second burst of activity during the swing phase of the cycle indicating hip flexion and ankle dorsiflexion, before GLUT and TA show an onset of activity in preparation for heel strike ahead of the TFL, PB, GL and GM muscles.

Gluteus Maximus activity was evident at heel strike as might be expected due to the hip stabilisation action of the muscle through support, and again during the propulsive phase as hip extension occurred. As the leg entered the last 30% of the stride cycle, GLUT activity was again noted as the thigh decelerated, hence arresting the forward movement of the thigh and preparing the leg for heel strike. Tensor fascia latae activity was evident to 13% of the stride cycle after heel strike, as might be expected, as it stabilised the hip for heel strike. As the thigh reached maximum extension, the TFL was again activated to flex the hip and accelerate the thigh in the direction of travel.

Both heads of the gastrocnemius muscle showed activity during the stance phase, as the muscle aimed to stabilise the ankle and decelerate the tibia as it passed over the ankle joint complex (MacIntrye and Robertson, 1987). Although the gastrocnemius also affects motion at the knee, concentric action was not required in this function during the support phase. The propulsion stage of stance showed continued gastrocnemius activity with the longest duration of activity in that stage, possibly suggesting its use as a prime mover in this stage. Both heads show a latency during swing, before the larger (medial) head contracts approximately 3% of stride ahead of the lateral head.

A similar pattern was observable for the peroneus brevis muscle trace. Activity during the stance phase could be attributed to a stabilising function at heel strike in the medio-lateral direction, with a plantar flexion action to aid the primary propulsion during the final stage of the stance phase.

The tibialis anterior was active earlier than the other ankle compartment muscles at approximately 75% of the stride cycle, causing dorsiflexion of the ankle in preparation for heel strike. This muscle also exhibited a long burst during the swing phase (middle burst in figure 3.6) after a latency of only 7.4% of the cycle, with swing phase activity to raise the toes after propulsion.

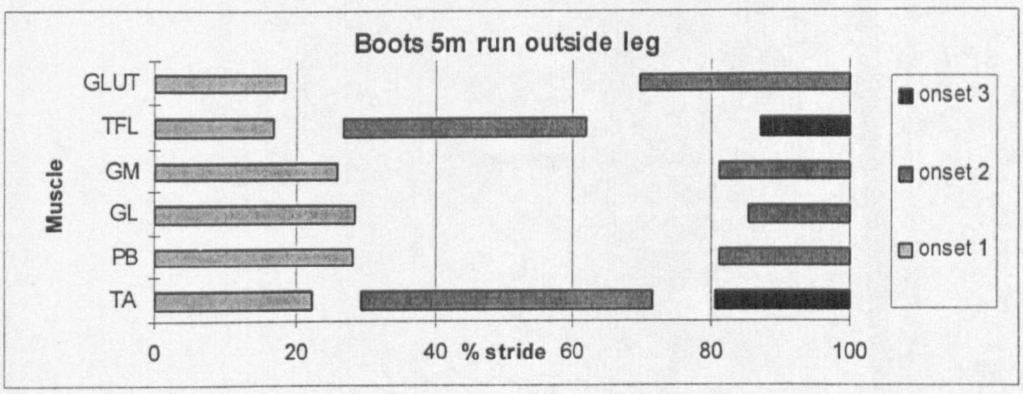

Figure 3.7 Mean temporal values for the outside leg running in boots at 5m radius (10 subjects).

Figure 3.7 shows outside leg activity at the 5m radius also differed from straight running patterns, with increased activity in most muscles for the outside leg. Onset 1 was longer than in both the straight running and for the inside leg at 5m radius (figure 3.8).

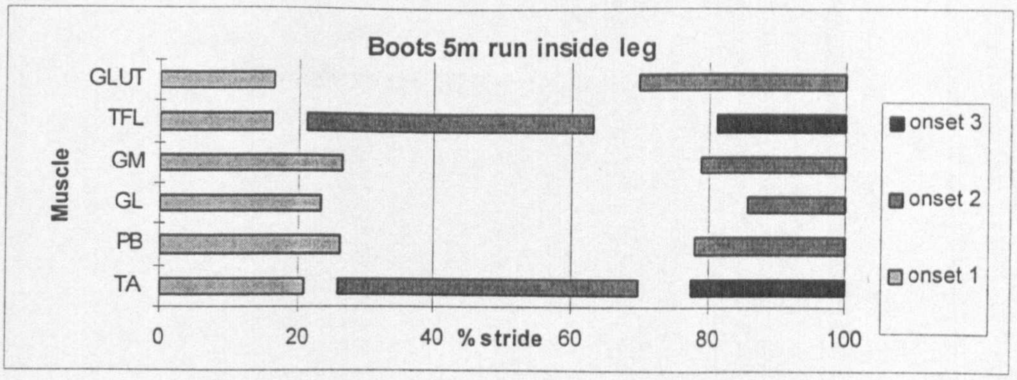

Figure 3.8 Mean temporal values for the inside leg running in boots at 5m radius (10 subjects).

Mean temporal values in the 5m run inside leg show an altered pattern from the straight run (Fig. 3.8). All activity bursts were longer in relative duration, resulting from earlier activation and later cessation through the cycle. The inside leg appeared to display a more pronounced increase in swing phase activity. Such altered temporal activity would suggest muscular adaptation to the different form of locomotion, with possible increased energy requirements for curvilinear running patterns. Although adaptation occurred at both 15m and 10m grades, differences were more apparent in the 5m condition.

Table 3.7 shows a stride kinematic analysis with stride length calculated from subject velocity and stride frequency (cadence) data obtained from footswitches. Stride length was significantly lower ($P<0.001$) in running at the 5m radius than in the straight trial, and velocity was not significantly different.

	JOG RADIUS				RUN RADIUS			
	0m	15m	10m	5m	0m	15m	10m	5m
velocity. ms^{-1}	4.56 ±0.105	4.36 ±0.010	4.52 ±0.010	4.37 ±0.011	5.54 ±0.168	5.48 ±0.010	5.48 ±0.012	5.10 ±0.009
frequency. s^{-1}	1.46 ±0.028	1.48 ±0.002	1.48 ±0.002	1.51 ±0.002	1.63 ±0.04	1.65 ±0.003	1.63 ±0.003	1.70 ±0.004
length m	3.18 ±0.093	2.98 ±0.007	3.08 ±0.007	2.93 ±0.008	3.47 ±0.11	3.33 ±0.008	3.39 ±0.009	3.00 ±0.007

Table 3.8 Mean kinematic variables and standard errors at differing curvilinear grades while wearing soccer boots.

The EMG recording analysis showed temporal muscle activity patterns for curvilinear motion which were inherently different to those of straight running (Table 3.8) or jogging (Table 3.9). Significant differences in the tables are represented by ☆ $P \leq 0.001$; ② $P \leq 0.02$; ④ $P \leq 0.04$; ⑤ $P \leq 0.05$. Duration at stance is represented by DUS.

	Inside leg muscle						Outside leg muscle					
TEMPORAL RUN EVENT	GLUT	TFL	GM	GL	PB	TA	GLUT	TFL	GM	GL	PB	TA
Offset 1 later at 5m			☆	☆	④		②	☆	☆	☆	④	☆
DUS greater at 5m		⑤					⑤	⑤			⑤	
DUS 0<15,10<5			⑤						⑤	⑤		
DUS 0, 15<10, 5					⑤							
DUS greater at 15m	⑤											
Onset 2 earlier at 5m		②										⑤
Duration 2 greater at 5m		④						④				
Onset final pulse earlier in curved											②	⑤

Table 3.9: Temporal differences in EMG activity when running on straight (0m), and curvilinear paths of 15m, 10m, 5m radius in soccer boots

Figure 3.8 showed activity in the GM displayed major adaptation to the grade of curve in the inside and outside leg. The first offset was later at 5m than other conditions, whilst the duration at stance was greater at 5m than the straight condition. As the SL was shorter in the 5m trials, body weight may remain more directly over the supporting foot. If such a change in body position means that the foot contact time increases, this could explain the adaptation. PB patterns exhibited similar adaptations to curved motion for both legs, with the offset of the first pulse occurring later in the

5m trials, and the duration at stance increasing for the tighter curved runs. The outside leg displayed further evidence of a change of PB activity as the final onset was later in the straight condition than curved, and was accompanied by similar changes in TA activity. Such observation may be explained by the reduced requirement for lateral stability of the ankle in the straight condition.

More adaptation to curvilinear motion was evident in the outside leg than inside leg, though in some cases the adaptations were the same e.g. GM. For all the muscles monitored two notable adaptations were evident; (i) At 5m radius muscles were active for longer at the beginning of the cycle, following heel strike i.e. PB and GM both legs, GL inside leg (ii) Greater duration at stance and swing at 5m in TFL.

	Inside leg muscle						Outside leg muscle					
TEMPORAL JOG EVENT	GLUT	TFL	GM	GL	PB	TA	GLUT	TFL	GM	GL	PB	TA
Offset 1 later at 5m than straight		☆	☆		④		⑤	☆	☆	☆	⑤	
Offset 1 later in 5m												☆
DUS greater in 5m										⑤		
DUS greater in 5m than straight								⑤			⑤	
DUS greater in curved							⑤		⑤			
Onset 2 earlier in 5m than straight		⑤										
Duration 2 longer in curved		⑤										
Offset 2 earlier in 15m						⑤						
Offset 2 0,15<10,5		⑤										

Table 3.10 Temporal differences in EMG activity when jogging on straight (0m), and curvilinear paths of 15m, 10m, 5m radius in soccer boots.

When jogging, the duration of the swing phase pulse in TFL increased in the inside leg, whereas duration at stance increased in the outside leg. General activity at the studied hip muscles followed a pattern of delayed offset of the first pulse in the outside leg at 5m.

Temporal adaptations to increasing curvilinear motion were evident but were most significant at the 5m grade of curve in both jogging and running. Following heel strike in jogging temporal muscular activity in the outside leg increased in stance and in the inside leg muscular activity increased in swing (Table 3.9). During running all muscles monitored for both the inside and outside legs showed adaptive mechanisms (Table 3.8). For jogging and running the majority of muscular adaptations to curvilinear motion occurred in the outside leg.

For soccer boots, adaptations to curvilinear motion appear to be similar for both the run and the jog conditions (Tables 3.8 & 3.9). However, in the jog condition, most muscles show modification between 5m and straight with less noticeable changes at the greater curve radii, but in running, most muscles show modification between 5m and all other conditions. The majority of those modifications to curvilinear motion occur in the outside leg of the curve. Consistencies between data sets that occur in both legs are centred on the GM and PB muscles:-

Both legs:	GM, PB -	later first offset after heel strike in 5m
Outside leg:	GLUT, TFL,GM,GL,PB,TA	" " " " " " " "
	GM, GL	DUS greater in curved
Inside leg: -	TFL	earlier onset and duration in swing.

The consistencies between the data sets highlighted show the modification to curvilinear motion in soccer boots. The significant differences in the two data sets of jogging and running, however, were most evident in the 5m condition. Temporal values in the 5m trials tended to differ from all trials in the run condition, but only differ from the straight trial in the jog condition. Such a pattern could be attributable to 5m being the most severe curve reproducible at the run speed, therefore requiring strong adaptations from the musculature. The same grade of curve at a jogging speed placed lesser demands on the musculo-skeletal system, resulting in more subtle changes.

Temporal muscle activation values were used as the prime analysis variable. Adaptations in these values suggested an altered muscle activity pattern during curvilinear motion. To justify the use of temporal EMG values in the analysis, the magnitude of muscular activity was considered. Amplitude of the EMG signals during straight and 5m trials were compared to elicit any differences in the magnitude of muscular activity.

	5m jog inside	5m jog outside	straight jog	5m run inside	5m run outside	straight run
TA st	62.9±4.0	62.1±4.3	69.7±2.3	64±3.2	57.5±5.0	68.7±3.7
TA sw	58.8±3.7*	57.8±5.2	57.4±3.0	69.5±3.5*	65.6±5.9	64.7±3.9
PB	88.9±13.0	105±11.8	102.2±8.8	86.4±12.4	108±11.6	104.4±8.0
GL	93.6±6.4	95.4±6.1	98.9±8.6	88.3±7.4	98.2±8.1	98.2±7.3
GM	101±7.9	100.3±7.6	101.4±8.0	97.4±8.4	96.2±9.7	103.1±9.4
TFL st	42.6±6.9	47.7±8.4	45.5±6.6	43.4±6.4	53.9±8.0	48.6±7.5
TFL sw	42.9±5.5	37.5±6.6*	38.7±5.3	52.8±6.9	56.1±7.5*	48.7±5.2
GLUT	78.3±10.4	81.8±14.7*	84.9±18.3	94.7±12.0	114±15.8*	108±22.8

Table 3.11 Mean ± S.E maximal enveloped EMG expressed as a % MVC for curvilinear motion in soccer boots. (st = stance; sw = swing, *P<0.05)

Mean maximal TA swing activity in running was greater than in jogging which was significant (P<0.05) in the inside leg. For the outside leg the mean maximal TFL swing activity and mean maximal GLUT activity were greater in running than jogging. These differences for the GLUT and TFL (swing) were significant (P<0.05). When all muscles were considered, significantly greater maximal enveloped EMG levels were recorded during running than during jogging (P=0.029). No significant differences between straight and motion at tightest curvature were found for either the inside or outside leg. As no differences in amplitude existed between straight and curvilinear motion, the use of temporal EMG as the prime analysis variable was substantiated.

Comparison of different conditions in jogging and running showed greater levels of TFL (sw) and GLUT muscular activity in running. When compared to temporal EMG results, these data accompanied increased duration of activity (Tables 3.8 and 3.9) in the outside leg during running. Both these muscles could act to provide greater abduction of the outside leg when running in a tight curvilinear path. Increased abduction during stance would move the body's centre of gravity towards the centre of the curve to maintain the direction of travel. Increases in swing phase TFL activity may arise from abduction and flexion at the hip and the need to cover greater distances with the outside limb. This finding would suggest that improved performance from these muscles in this type of activity might be an asset. Training of these muscles by repeated performance of tight curvilinear running, or hip abduction and flexion/extension resistance training could be of benefit.

In summary, whilst wearing soccer boots a general adaptation of temporal activity was noted in all monitored muscles at the tightest curvature within a right heel strike to right heel strike stride. Muscle EMG amplitudes were generally greater during running than jogging, and GLUT and TFL muscles demonstrated altered magnitude coinciding with altered temporal activity. Performance of curvilinear motion did not appear to alter the magnitude of muscular activity in either jogging or running. In all muscles altered temporal activity coincided with a tendency towards reduction in velocity and stride length and an increase in stride frequency (Run: velocity 7.9%, stride frequency 4.2%, stride length 13.5%; Jog: velocity 4.2%, stride frequency 3.3%, stride length 7.9%) during motion at the tightest curvature.

3.3.3

COMPARISON OF BOOTS AND FLAT SOLED SHOES

Boots and Shoes in Linear Motion

When the two graphical outputs of mean temporal data for straight running in shoes and boots were compared, a very similar pattern was evident. From figures 3.2 and 3.6 the main differences appear in the PB muscle. During running in boots the initial offset were approximately 2.7 % later and the final onset 4.8 % later than corresponding shoe trials. However, in both heads of the gastrocnemius muscle, running in boots produced an earlier activation pattern with a difference of approximately 2 % offset and 2.5 % onset.

Differing frictional qualities at the shoe-surface interface were expected to alter the temporal EMG activity pattern. However, all other monitored muscles displayed similar patterns in both types of footwear, suggesting no fundamental changes in temporal muscle patterns with different frictional qualities during straight running. Absence of differing temporal activity could be explained by limiting friction in straight motion not reaching levels of friction provided in either of the footwear conditions, therefore the altered frictional characteristics were not of sufficient magnitude to elicit EMG changes. Stride length was greater during straight running in shoes, but could be attributed to the slightly higher mean velocity. Magnitudes of muscular contractions were significantly ($P < 0.05$) higher whilst running than jogging, but were not affected by altering footwear, and are hardly surprising.

Boots and Shoes in Curvilinear Motion

When comparing the differences in temporal activity in footwear types during curvilinear motion it would appear logical the greatest differences would be elicited at the tightest 5m radius. Figures 3.3 and 3.7 depict temporal activity in the outside leg of the 5m curve. TA patterns were similar in each condition, whereas PB shows a delay in initial offset and final onset in the soccer boot condition, similar to that of straight running. A lesser delay in activity was evident in both gastrocnemius heads and GLUT activity. TFL activity altered by increasing the duration at stance when shoes were worn, and increasing duration in swing when soccer boots were worn, possibly a reaction to the decreased friction experienced in the shoes condition.

Graphs representing inside leg activity at the 5m radius are shown in figures 3.4 and 3.8 for shoes and boots respectively. Slightly differing patterns of activity were demonstrated in all muscles monitored. The main consistency evident from the graphs is the earlier cessation of GL in relation to the other muscles than in either the straight or outside leg activity, suggesting the main force through the Achilles tendon is created by the medial side of the gastrocnemius whilst the lateral side of the foot remains in contact with the ground. Qualitative evidence from video recordings suggests subjects may not make heel contact with their inside leg whilst performing at the tightest grade of curve, a feature of the gait pattern which may give further insight into the altered activity shown if investigated in greater detail. GLUT activity remains approximately equal in both footwear conditions with a slight increase in duration at stance in soccer boots. Swing phase activity of the TFL shows a large increase in soccer boots, possibly indicating more abduction and flexion in boots. Such a result could be explained by the increased friction in the boots providing a more solid base for ground reaction to such actions. TFL also showed a shorter duration at stance in the shoes condition. GM, GL and PB muscles showed similar patterns after heel strike yet GM and PB activated earlier in preparation for heel strike in the boots condition. As GL appears to cease activity prior to GM and PB it is possible that GM and PB may be working antagonistically to replace the foot in a neutral position. TA demonstrated a similar pattern in both footwear types with swing phase activity occurring slightly earlier in soccer boots.

Differences described in this section represent the subtle adaptations to temporal muscle activity patterns when the friction at the shoe-surface interface was altered during curvilinear motion. Although such findings were averaged from results of ten subjects it remains difficult to speculate on the reason for such altered activity in the muscles monitored. As all muscles control movement outside of the saggital plane it is possible that such differences serve to maintain balance when different frictional properties between the athlete and the sports surface occur. A general pattern appears to be a shift of activity in selected muscles to later in the stride cycle whilst wearing soccer boots. A possible explanation to these results could be the foot position required before contact. When wearing soccer boots and running in a curved pattern, one side of the boot will penetrate the turf more than the corresponding side. Such a notion would suggest that when running a curve in soccer boots one would not have to pronate/supinate the foot a great amount prior to foot contact. However, in training shoes, no turf penetration occurs, meaning the foot must be aligned closer to a foot-flat position before contact. Such preparation would require increased pronation/supination of the foot and may explain the earlier muscle activity in these trials.

No kinematic changes were noted between shoes and boots, although overall pooled kinematic results followed the pattern of decreased stride length and increased cadence at the 5m radius ($P < 0.001$). Whilst player speed also showed significant differences, these were only apparent at the 5m radius ($P = 0.001$). Such results may undermine the kinematic changes shown at that grade of curve, but should not effect the adaptations in temporal muscle activity as all data were normalised to percentage of stride.

Patterns in Outliers

Triple burst PB activity was displayed in one subject compared to the more general single burst of activity (see figure 3.9). At 5m curvature the inside leg PB showed no activity at heel strike in two subjects though normal PB activity was shown in the outside leg condition, possibly indicating a minority adaptive mechanism to curvilinear motion. Accompanying the absence of activity at heel strike appeared to be a general delay of PB activity throughout the cycle in these conditions.

With those subjects showing three discrete bursts of activity in the peroneus brevis, the mean muscle pattern showed a slightly earlier onset. The second and third pulses would show an overlap with the final bout of activity in those subjects exhibiting single or double bursts of muscle activity.

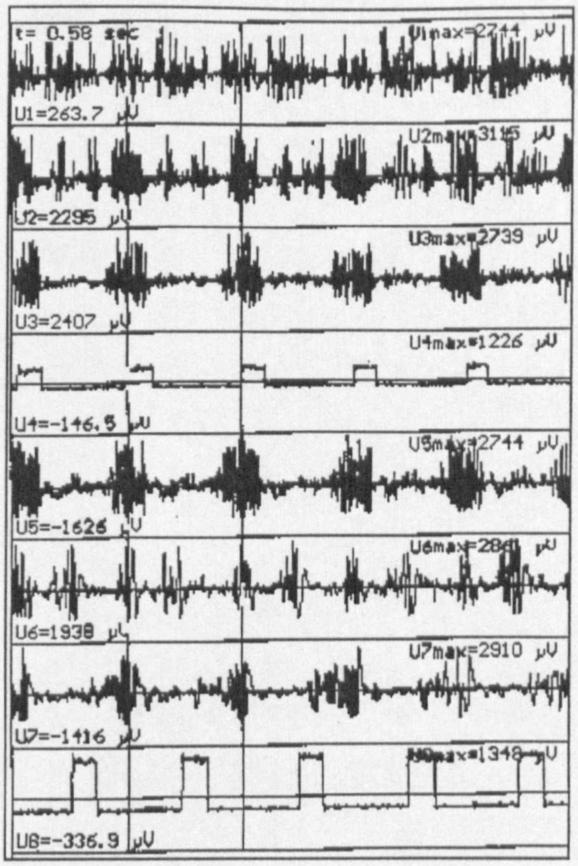

Figure 3.9 Triple burst PB activity in channel 2.

Three subjects exhibited swing phase activity in the lateral head of the gastrocnemius. Two subjects showed this activity in approximately half of their trials, whilst another subject only displayed the feature in four trials. However, for those subjects exhibiting the pattern, adaptation was evident when comparing inside and outside leg activity, with occurrence less frequent in the outside leg. This may be due to the preparation for a more pronounced front foot strike in the inside leg, especially in the more severe curves, placing increased strain on the gastrocnemius. All subjects claimed to be injury free at the time of testing, but no assessment of biomechanical abnormalities was made.

Although such activity was only evident in a minority of subjects, adaptation to curvilinear motion was noticeable. One subject showed the extra burst of activity alternated from just before stance to just after stance, dependent on whether the inside or outside leg was monitored.

A deviation from the standard single pulse medial head gastrocnemius function was also noted in one subject. Activation in preparation for heel strike occurred earlier in the cycle, and would cease between 8% and 18 % before heel strike, only to recommence at heel impact with the turf. This type of gastrocnemius activity occurred only in the outer leg for five trials, and also in the 'boots straight jog' condition. This subject would appear to exhibit an opposite effect to those displaying a swing phase pulse for the GL muscle. As the medial head lies on the opposite side of the calf, its activation for trials using the outside leg would be logical. Such activity would aid foot placement requiring increased inversion as the body leans into the turn.

Hypotheses

1) Temporal muscle activity showed adaptation to curvilinear running ($P < 0.05$) when compared to linear running, therefore the null hypothesis is rejected.

2) Temporal muscle activity in the gluteus maximus, tensor fascia latae, gastrocnemius, peroneus brevis, and tibialis anterior showed statistically significant increased adaptation as the grade of curve became more severe, therefore the null hypothesis is rejected.

3) The magnitude of muscular activity as measured by EMG amplitude did not display adaptation from linear to increased severity of curvilinear motion, therefore the null hypothesis is retained.

4) The amount of muscular activity around the stance phase was reduced in soccer boots as opposed to training shoes, therefore the null hypothesis is rejected.

3.4

GENERAL DISCUSSION

The results presented display a temporal EMG muscle activity pattern for curvilinear motion which is inherently different to that of straight running or jogging. The results also highlight the effect of differing frictional properties at the shoe-surface interface in curvilinear performance.

General Activity Patterns in Running/Jogging

Comparisons of common muscles from the present study with those available on linear gait in training shoes was possible, though alternative patterns were evident. The present study found that during the conditions of linear gait the gastrocnemius showed activity at approximately 85% of the stride cycle. These findings would agree with those of Mann and Hagy (1980a) and Nilsson et al. (1985). In comparison, Schwab et al. (1983) noted that calf muscles were activated just before initial contact, at 98% of the cycle. However, the mean velocity used was low (2.45 m/s). Schwab et al. (1983) claimed that midswing calf activity consistently appeared in their investigation, but the authors stated this could not be quantified, perhaps due to its occurrence in only a small number of subjects, as in the present experiment. Midswing calf activity was also reported by Mann and Hagy (1980) at 5.3 m/s, but not by other authors.

Data from the present investigation indicated an earlier onset of the medial head of the gastrocnemius compared to the lateral head, prior to heel strike. Such activity would appear reasonable considering the movement occurring in this phase. The tibialis anterior (TA) contracts to raise the toes in preparation for shock attenuation at heel strike. The observed pattern of activity is explained as the TA also causes inversion of the foot allowing contact on the lateral border of the heel. This action would require shortening of the medial head of the gastrocnemius more than the lateral head. Alternatively, the earlier onset may be explained by the neuromuscular control of the larger medial head possibly dominating the eccentric contraction during this action, a suggestion that demands further inquiry.

Consistent muscle patterns were displayed by the tibialis anterior. Timing of muscle onset and cessation varied only slightly across all trials, whilst the number of bursts of activity per cycle remained constant at all grades of curve. Kameyama et al. (1990) investigated EMG patterns around the ankle joint in Japanese subjects. The authors reported different patterns for tibialis anterior of double burst, triple burst and continuous firing in straight running. The present study of Caucasian subjects shows only a double burst of activity. However, if the raw EMG traces are examined carefully, some subjects appear to show a tendency for division of the second burst. Such activity saw the amplitude of the signal oscillate but never drop to a level approaching that which could have been considered as non-activity, and therefore cessation of that discrete burst. Therefore, for those trials demonstrating varying amplitude in a single burst, temporal values recorded were those encompassing the whole burst. The signals presented in Kameyama et al. (1990) clearly distinguished between the patterns of firing, with the main differences occurring in the stance phase. Unfortunately no comparison to Caucasian subjects were made and detailed methods of data collection were not provided, so differences in muscle pattern cannot be attributed to either the population sample, or the method of data collection.

In all traces the TFL displayed a double burst of activity. The TFL has been subject to little relevant study in the scientific literature to date, with the notable exception of Pare et al. (1981), who attempted to differentiate the function of anterior and posteriolateral fibres using fine wire bipolar electrodes. They discovered that the two compartments of the muscle demonstrated conflicting functions, with the posteriolateral fibres being responsible for inward rotation and abduction of the thigh whilst the anterior fibres flexed the thigh. From the traces provided in the results section it would appear that both functions have been recorded although the posteriolateral portion of the muscle was targeted. The synchronisation of these pulses was checked by reference to videotape. However, the use of surface electrodes meant the signal had become representative of a more holistic muscle action. Possibly a more representative measure of non-linear motion could be attained if only the single pulse around the stance phase was considered for variations in temporal values. This would enable identification of temporal differences as the muscle functions to medially rotate and abduct the thigh, whereas swing phase activity would appear to primarily represent hip flexion. In the present study, both stance and swing phase pulses were analysed. The analysis of both pulses yielded some interesting findings from the swing phase pulse in the outside leg whilst wearing soccer boots. It was thought such results justified the inclusion of the data in the study.

Temporal variables enable the subtle timing differences to be identified that distinguish between different paths of gait. However, if a number of muscles which fulfil related functions at a joint are active over a similar period, it becomes difficult to specify the muscle designated as prime mover if signal amplitudes are not considered. For example, in the propulsive phase of the cycle Elliot and Blanksby (1979) reported the magnitude of triceps surae activity would support their role as the prime movers in this phase when compared to amplitudes of the other muscles monitored. Such claims would however infer that all muscles relevant to the considered action were monitored. Data from the present study showed alterations in muscle amplitudes only coincided with temporal differences when the subjects were wearing soccer boots (Tables 3.8, 3.9 & 3.10). Such results suggested that TFL (sw) and GLUT muscles could be the key muscles in the outside leg during curvilinear motion at a 5m radius but could not be considered as the prime movers for such actions as no recordings were made from muscles such as ileopsoas or hamstring group. The inclusion of these muscles in the present study would have meant the loss of data from key muscles studied. However, these additional muscle groups could be investigated in future studies by systematic rotation of the electrodes.

The present study

The kinematic measures in the present study show two major changes in the performance of the tighter 5m grade - a decrease in stride length and an increase in stride frequency; these modifications in kinematics are required to aid performance. Shortened stride length would maintain the body in a more upright posture and promote balance. If foot contact time remained constant in these actions the athlete would increase the control of the movement due to an increased percentage of foot contact, and the adaptations to curvilinear motion described from the temporal EMG data could be a function of increased percentage foot contact. Such knowledge is, at present unavailable, yet vital if the reasons for increased muscular activity around the stance phase are to be understood. This hypothesis appears worthy of further investigation.

Curvilinear motion performance prolonged activity in all muscles of the outside leg after heelstrike and increased the duration of activity in the swing phase of TA and TFL. Increased muscle activity around stance would enable the prime muscles for curvilinear progression increased time to position body segments. Although EMG magnitude was not significantly different between curved and straight trials, the increased time of application enables greater muscular impulse to be applied due to the proportionally increased foot contact time.

Major adaptations were only evident at the 5m radius although changes to temporal activity were shown to a lesser extent at the other radii. Although the increase in the grade of curve was constant from 15, to 10, to 5m, the proportional increase in curve severity was greater towards the smaller radii. Such non-proportional increase in the severity of curve could go some way towards explaining why adaptations were mainly evident in the 5m trials.

Video records for the 5m trials show that subjects alter foot contact in the 5m condition towards a forefoot strike, especially in the inside leg. Adaptation of technique in such a manner could be caused by increased activity in the tricep surae muscle group. Such a change in technique would entail less dorsiflexion in the support phase of the cycle. However, such observations remained unsupported by temporal muscle activity in muscles of both the anterior and posterior sections of the lower leg. Compensatory movement for forefoot strike appeared to come from the torsional action of the foot at impact giving forefoot varus. Such action would maintain the contact area of the forefoot with the ground whilst the rearfoot remains in a neutral position relative to the shank. Such torsion would lead to reduced stress on the Achilles tendon complex as muscular force was applied.

Running in Boots

Muscular adaptations to curvilinear motion in the lower extremity can be identified when comparison is made with linear gait. In analysis of straight to curvilinear motion in running, the majority of differences in temporal activity were evident in the outside leg of the curve. The muscular force from the tangential component of motion in a curved path would increase as the grade of curve became tighter. The muscular force applied to the outside limb would also be more eccentric to the body centre of mass than the inside limb, due to body orientation (lean into the curve). A combination of these phenomena could explain the greater level of adaptation occurring in the outside limb.

The later final onset of activity prior to heel strike in the TA and PB in straight running in boots can be attributed to the reduced requirement for lateral stability when compared to curved motion. All other major adaptations occurred at the 5m radius, therefore it must be noted that stride length was shortened at the 5m radius, placing the centre of gravity more directly above the supporting foot. However, as the grade of curvature becomes tighter, the degree of body 'lean' should also increase. Such an action would serve to reduce the proportion of body weight acting normally to the ground. Therefore, a greater contact period may be required to generate the required mediolateral impulse required for curvilinear motion (see diagram in chapter 1). Such altered stride kinematics could explain the increased muscular activity around the support phase at 5m. The main findings for running in boots were a longer phase of activity at the beginning of the stride cycle, following heel strike ie. PB and GM in both legs, and GL inside leg. Also a greater duration at stance and swing at 5m in TFL.

Jogging in Boots

The adaptations to curvilinear jogging in boots followed a similar pattern to those in running, as the outside leg displayed more changes. The majority of adaptations again occurred around the support phase (Table 3.9). Temporal differences were noticed between 5m radius and straight motion, whereas in running changes were observed between 5m and all other grades of curve. Due to the reduced velocity in the jogging trials, the muscular force exerted was less. Thus, differences in temporal values appear only between the extremes of straight and 5m curve.

Footwear differences

If greater stability and control are required for tight curvilinear performance, and gained from altered motion kinematics, it would be reasonable to suggest that a reduced frictional value in the use of flat soled shoes would also illicit a kinematic change. A greater proportion of the stride cycle spent in the contact phase ought to redress the loss in friction in the flat soled condition. However, no kinematic differences were observed between boots and shoes in straight motion, but TA, PB, and TFL inside leg values showed increased duration of activity at stance when jogging in shoes. TFL inside leg duration at stance was also increased for the shoe condition in running.

If a footwear effect were to exist during curvilinear motion performance, systematic patterns would be expected in the data. At the 5m radius when running, the pattern was for an earlier onset of TFL in the swing phase pulse of the boot conditions. One would not expect such a result to be a consequence of changing frictional characteristics, as the limb does not contact the ground in the swing phase. However, as the boots will provide more friction, this creates a more stable base for the earlier abduction of the inside leg in preparation for the next footstrike. Such a situation should provide a smoother turning action in soccer boots than in shoes. The increased abduction of the inside leg during swing would also suggest a capability for tighter turns in soccer boots than in flat-soled shoes.

The test protocol employed enabled the satisfactory performance of trials in all conditions. Therefore, the frictional changes in the two conditions were perhaps not sufficiently different to elicit a more marked adaptation. However, the performance in the flat soled shoe conditions was not possible using a faster running velocity or a tighter radius of curve.

Jogging in flat soled shoes displayed a unique pattern which would suggest an adaptation at the muscular level not connected to altered stride kinematics. The final onset of activity in the TFL and PB muscles was earlier in the 5m condition, which contrasts with results of the other data sets. Modification to curvilinear motion had previously followed a pattern of a shift to later in the stride cycle. The duration at stance for 5m was increased by later offset of the first pulse in other data sets, whereas in the jog with flat-soled shoes this period increased by an earlier activation prior to heel strike. Such a pattern could not be connected to an increased percentage stance phase, as the increased activation occurred prior to heel strike. Possible explanations could not be attributed to either changing velocity or shoe-surface frictional characteristics, as these results were not apparent when comparing jogging in boots or running in shoes. The resulting muscular adaptation must therefore arise due to a complex interaction between the altered velocity and footwear type.

3.5

CONCLUSIONS

Differences in temporal EMG patterns were evident between straight and non-linear motion, and occurred primarily at the 5m radius of curve. This was due to the greater proportional increase in curve severity for these trials. Adaptation occurred in both legs, although predominantly in the outside leg of the curve, with the general form of increased duration of muscular activity after heel strike. Differences in muscle adaptation occurred with subject velocity. When running, differences were generally noted between values at 5m and all other trials. However, when jogging, values only varied significantly between 5m and straight trails. Such results suggest that during slower velocity trials, the difference between the grades of curve chosen was not sufficient to elicit a significant adaptation in temporal muscle activity patterns.

Muscles showed differing degrees of adaptation. The gastrocnemius medial head was a prime adapter, exhibiting differences at the 5m radius in boots and shoes at jogging and running velocities. All other targeted muscles displayed adaptation to curvilinear motion during some conditions. Those muscles showing consistent adaptation to curvilinear motion, irrelevant of running velocity or footwear type were:-

1. GM later offset of first pulse both legs, run and jog.
2. GLUT later offset of first pulse outside leg, run and jog.
3. TFL later offset of first pulse outside leg, run and jog.
4. TA later offset of first pulse outside leg, run and jog.
5. PB later offset of first pulse inside leg, run and jog.

The results presented form a pattern of muscle activity different to that observed in straight (linear) motion and may form the basis for mechanisms of specific movement patterns required for successful soccer performance. However, it is impossible from results to distinguish between increased duration of muscular activity, and the mechanism of an increase in muscular torque applied to the ground. A small proportion (15%) of tests showed irregular muscle activity patterns in the muscles tested, which may be due to functional differences of these subjects, or may have arisen due to injury related limitations. Although some patterns did not match average traces from other subjects, adaptations to curvilinear motion were still evident.

Temporal muscle activity was altered from straight to curvilinear motion as were stride kinematics. Increased activity typically occurred around the stance phase of the stride cycle. However, as only heel strike was identified during EMG investigations it has been impossible to suggest that the altered muscular activity occurred within an otherwise identical stance phase. If the time taken in the stance phase had altered during curvilinear motion, differing values of temporal EMG could be at least partly explained as mechanisms to alter paths of the limbs. Alterations in stride length and frequency suggest that kinematics are altered to increase the stance phase, and results of temporal muscle activity show consistent increases at that phase. Therefore, results were required to explain the relationship between foot contact time and curvilinear motion. The following study reports an investigation to determine foot contact time during curvilinear motion, in an attempt to discover possible explanations for increased duration of muscular activity around the stance phase.

CHAPTER 4

CHAPTER 4

An Investigation into Foot Contact Time and Stride Kinematics During Curvilinear Motion

4.1

INTRODUCTION

The previous study into curvilinear motion showed adaptation in temporal values of onset and cessation of muscular activity. Accompanying the muscular adaptation, stride kinematics were modified with a trend towards reduced stride length and increased stride frequency as the grade of curve became tighter. In an attempt to explain the temporal muscle adaptations around stance, it is essential to gain knowledge of foot contact time during curvilinear motion at different grades of curve.

As straight running speed increases, times of support and non-support have shown to decrease (Nelson and Osterhoudt, 1971; Nelson et al., 1972). However, little has been presented concerning kinematic alterations in non-linear motion. For curvilinear motion Greene and McMahon (1979) measured subjects running at maximum speed along circular arcs of different radii. Top speed, ballistic air time, and ground contact time were reported to change dramatically with radius. Greene and McMahon (1979) also claimed that neither step length nor frequency altered appreciably as a function of the radius, and therefore were assumed constant for each subject and essentially independent of the radius. Such data would appear to conflict with the measures of stride kinematics presented in chapter 3, where stride length and frequency were shown to alter as the grade of curve became tighter. However, in a subsequent investigation, Greene (1985) reported stride length to decrease exponentially with a tighter radius, in agreement with findings by the author of this thesis, presented in chapter 3.

From the studies in chapter 3 it was evident the majority of muscular adaptation occurred around the stance phase of the stride cycle. All of the recorded muscles perform a stabilisation function as part of their activity during stance, in addition to their role of accelerating body segments outside of the sagittal plane. Increased stance time as the grade of curve became tighter would explain much of the increased temporal activity reported. Even if actual foot contact time remained constant at all grades of curve, the proportion of the stride cycle spent in contact with the ground would increase due to the increased stride frequency. If such an assumption was correct, then a prime adaptation to curvilinear motion would appear to be altered stride kinematics. The altered temporal muscle activity would then serve to maintain stabilisation of the lower extremity during stance. This stability could then allow for the body 'lean' associated with curvilinear performance. Body 'lean' would then allow for the application of mediolateral force necessary to maintain the centripetal force required for motion in a curved path. Therefore, knowledge of the foot contact time during curvilinear motion would aid understanding of the biomechanical mechanisms working to create curvilinear motion. Data would also redress the conflict present in previous studies (Greene and McMahon, 1979; Greene, 1985). The research study reported in this chapter investigated the effect of curvilinear motion upon outside and inside foot contact time. From feasibility, pilot and main study experimentation reported in chapter 3, preliminary evidence suggested an increased tendency towards greater foot contact as the radius of the curve become tighter, especially in the outside leg. In addition a purpose of this study was to investigate the division of total foot, rearfoot, and forefoot contact times as visual observation during the experimental study reported in chapter 3 indicated that there was possibly a change in the forefoot-rearfoot contact time relationship.

Hypotheses

H_1: Inside leg total foot contact time will increase with severity of radius.

H_2: Outside leg total foot contact time will increase with severity of radius.

H_3: Inside leg forefoot contact time will increase with severity of radius.

H_4: Outside leg forefoot contact time will increase with severity of radius.

H_5: Inside leg rearfoot contact time will decrease with severity of radius.

H_6: Outside leg rearfoot contact time will decrease with severity of radius.

4.2

METHOD

Subjects

Eight male subjects familiar with curvilinear soccer movements volunteered for the study (Age 27.1 ± 4.7 years). Each subject had foot size equivalent to UK size 9 to enable standardisation of footwear. Each player wore new standard six-stud soccer footwear (Mizuno, pro-model UK size 9). All subjects were in good health at the time of testing. Ethical clearance and informed consent were obtained and each subject was reminded of their right to withdraw from the study at any time. Environmental conditions were the same as in chapter 3, the difference with respect to subjects was N=8 in the present study, compared with N=10 in chapter 3.

Instrumentation

Foot contact time was measured using two footswitches positioned inside the heel section and under the metatarsal heads of the right boot in a polyurethane insole. The switches were connected to an eight channel radio telemetry system (MIE Research Ltd. MTR8; Leeds, England) and a yagi aerial transmitted the data from the free roaming subject. Data were sampled as a DC signal from two channels at 500Hz and recorded on a Viglen 4DX33 personal computer running Orthodata GmbH MYO-DAT 3.0 software for MIE MT8-MBM.

Procedure

To enable comparison with data from chapter 3, a similar protocol was employed. Radii were measured at 5, 10 and 15 metres. Jogging and running velocities remained the same as chapter 3. Data were sampled over a five second period. As data were sampled for the right leg, the direction of travel around the curve was again reversed for each condition to enable data capture for both the inside and outside legs of the curve. For comparison subjects also completed jogging and running in a straight path.

Data Analysis

The overall period of the stride cycle was averaged from 6 - 8 strides, with stride length computed from the relation linking subject velocity to the product of stride length and stride frequency. From each trial three typical strides were identified by visual inspection of the data. For each stride, measurements of rearfoot contact, forefoot contact, and total foot contact time were taken by cursor movement in the Orthodata software. A three factor (grade, leg, speed) analysis of variance was performed within each subject to determine differences in the data.

<u>4.3</u>

RESULTS

Average results were calculated for the eight subjects and are presented below in tabular form.

Condition	Stride frequency s^{-1}	Stride length m
straight jog	1.44±0.02	3.09±0.05
15 outside jog	1.44±0.02	3.10±0.05
10 outside jog	1.41±0.02	3.14±0.04
5 outside jog	1.47±0.03	3.05±0.07
straight jog	1.44±0.02	3.09±0.05
15 inside jog	1.44±0.02	3.05±0.05
10 inside jog	1.42±0.02	3.18±0.05
5 inside jog	1.44±0.03	3.10±0.06
straight run	1.54±0.03	3.53±0.05
15 outside run	1.60±0.04	3.47±0.10
10 outside run	1.56±0.03	3.46±0.06
5 outside run	1.70±0.06	3.17±0.12
straight run	1.54±0.03	3.53±0.05
15 inside run	1.60±0.03	3.38±0.06
10 inside run	1.58±0.04	3.45±0.07
5 inside run	1.64±0.05	3.27±0.12

Table 4.1. Mean (±S.E) curvilinear jogging and running temporal parameters while wearing soccer boots (n = 8).

Subject velocity remained within 5% of the target values for each test, and was maintained throughout all grades of curve. Table 4.1 indicates that the transition from jogging to running was associated with significant increases in stride length for the inside and outside legs (P<0.001). In the outside leg during running mean values indicated there was a significant reduction in stride length from straight to curvilinear trials (P < 0.05), but stride length alterations in the inside leg were not significant.

Adaptation to curvilinear motion was evident at the tightest curvature with a reduction in stride length and an increase in stride frequency and was more clearly evident in running than jogging. Statistically, stride length was significantly reduced during running at the 5m radius (P<0.001) than at other grades of curvature and straight motion. However frequency displayed the greatest adaptation to curvilinear motion of the kinematic measures taken.

With adaptation occurring primarily at the 5m radius. Stride frequency was greatest in the outside leg during running at this smallest radius and differed between grade of curve (P=0.004), between inside and outside legs (P=0.035) and with speed of locomotion (P<0.001).

Condition	Outside leg s	Inside leg s
straight jog	0.31±0.01	0.31±0.01
15 jog	0.29±0.02	0.30±0.01
10 jog	0.31±0.02	0.30±0.01
5 jog	0.30±0.02	0.29±0.01
straight run	0.29±0.02	0.29±0.02
15 run	0.28±0.02	0.26±0.01
10 run	0.30±0.02	0.26±0.02
5 run	0.32±0.03	0.27±0.01

Table 4.2 Mean (± S.E.) duration of foot contact time (seconds) while wearing soccer boots (n = 8).

Contact time remained relatively consistent for both legs at all grades of curve when jogging (Table 4.2; Figure 4.1). Figure 4.1 shows that for the inside leg during jogging, relatively similar rearfoot and forefoot contact times were recorded with increasing curvature. In the outside leg as the curvature became tighter there was a significantly greater rearfoot contact time (P = 0.003) than in the inside leg. Although mean forefoot contact time decreased as the curvature became more severe in the outside leg, the trend was not significant statistically (P = 0.679). The division of foot contact time always showed the forefoot time to be longer than the heel contact time, except in the outside leg during jogging at the tightest curvature.

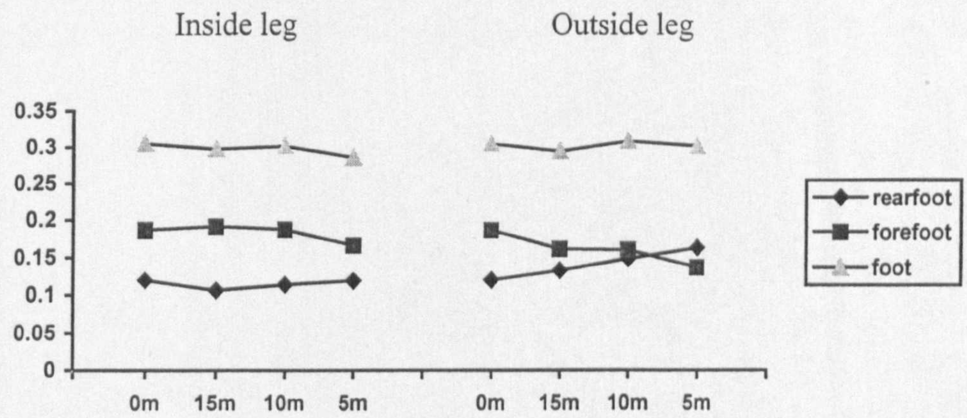

Figure 4.1 Relationship of heel, forefoot, and total foot contact time whilst jogging.

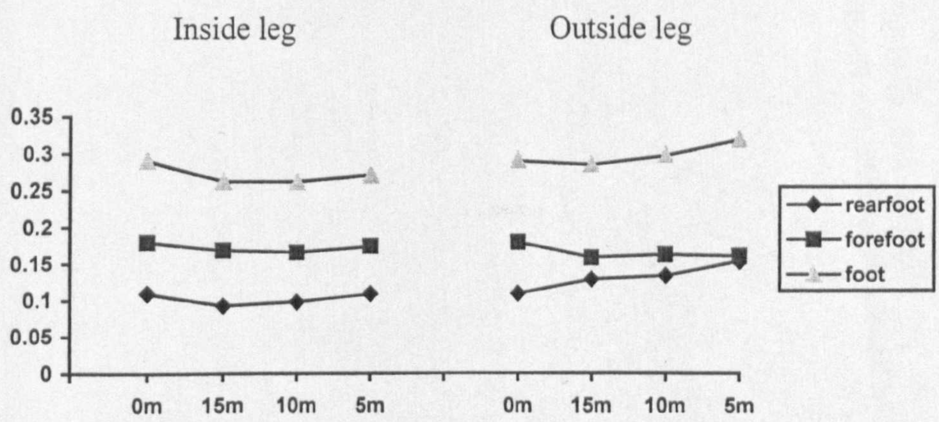

Figure 4.2 Relationship of heel, forefoot and total foot contact time with curvature whilst running.

When considering jogging and running the increase in movement velocity caused a significant overall reduction in rearfoot contact time (P = 0.035) (Fig 4.1 and 4.2). Figure 4.2 indicated that a similar adaptation was evident at the running velocity, where the inside leg showed similar rearfoot and forefoot contact times with increasing curvature. The outside leg showed a significantly greater rearfoot contact time than the inside leg (P = 0.003), although forefoot contact time did not differ between legs (P = 0.405). There was a strong trend for time of rearfoot contact to increase with increasing curve severity (P = 0.05). A significant interaction effect was observed (P = 0.003) for the increase of rearfoot contact time at the outside leg as the curve became tighter.

The results indicate that the only significant adaptation to tighter curvature (decreasing radius), was increased rearfoot contact time in the outside leg relative to the inside leg.

1) Total foot contact for the inside leg did not increase with severity of radius, therefore the null hypothesis is accepted.

2) Total foot contact for the outside leg did not increase with severity of radius, therefore the null hypothesis is accepted.

3) Forefoot contact time at the inside leg did not increase with severity of radius, therefore the null hypothesis is accepted.

4) Forefoot contact time at the inside leg did not increase with severity of radius, therefore the null hypothesis is accepted.

5) The inside leg rearfoot contact time did not increase with severity of radius, therefore the null hypothesis is accepted.

6) The outside leg rearfoot contact time increased with severity of radius, therefore the null hypothesis is rejected.

4.4

DISCUSSION

The results show that there was significant adaptation to curvilinear motion in jogging and running in both stride length and stride frequency, thus confirming the findings of earlier experiments (chapter 3). However, no significant adaptation was found with increasing curvature in total foot contact time, or forefoot contact time for the inside or outside leg. In the outside leg there was a significant increase in rearfoot contact time with severity of curvature.

The perceived anomalies observed between these data and chapter 3 may be explained by referring to the overall difference in stride period. The results presented from the presented study show that the foot contact time does not alter. However, as the stride kinematics tend to reduce the time spent performing each stride, the proportion of the stride cycle spent in contact with the ground will increase. That is, even though the total foot contact time does not increase, the proportional foot contact time does. This increase could go someway towards explaining the adaptations shown in temporal EMG patterns after heel strike in previous investigations (chapter 3). This proportional increase in foot contact time would seem to be a prime mechanism for curvilinear motion performance at a muscular level. With kinematic data now available to verify these claims, the adaptation of proportionally increased foot contact time was clearly elicited during analysis of temporal EMG patterns. Such patterns were evident in the greater duration of the first pulse of activity after heel strike and a greater duration of activity around the stance phase. (Section 3.5 Results; GM later offset in both legs; GLUT, TFL, TA later offset in the outside leg; PB later offset in inside leg)

The results presented above give rise to a conflict with previous studies in the literature of stride kinematics. Stride frequency was deemed independent of grade of curve by Greene and McMahon (1979) and Greene (1985) yet was shown to alter as a function of curve in the present investigation. An explanation for this could be that the velocity of running in the present study was maintained at discrete values of 4.4 and 5.4 m/s, whereas Greene and McMahon (1979) instructed subjects to run at maximal velocity. Running at maximal velocity might presumably require a maximal stride frequency, which would therefore be essentially independent of radius. Whilst Greene (1985) claimed step length and stride time to be deemed independent of radius, graphical data showed stride length to reduce exponentially as the radius decreases, which would form closer agreement with the present study. Greene and McMahon (1979) found ground contact time to range from approximately 0.12 to 0.20 seconds from 80 feet and 12 feet radius curvature respectively. The contact times in the present investigation range from 0.27 seconds during straight running to 0.32 seconds during running at 5m radius. The differences can be explained by the maximum velocities used by Greene and McMahon (1979) and Greene (1985).

The footswitch method was considered feasible during a pilot study, yet in practice it was difficult to attain the data. The researcher had to be very sensitive to experimental procedure due to the frailty of the footswitches during the main experimental study. A more robust method of investigating foot contact time was needed if a similar experiment were to be repeated. Using a force platform for example to monitor foot contact time would enable verification of results collected both here, and in previous investigations. The data presented here showed foot contact time did not alter as a function of curvilinear motion. Such results showed a proportional increase in foot contact time if the reduced stride frequency and period of stride were considered, especially at the tightest grade of curve. Future chapters will attempt to verify these data on foot contact time with the use of force platform analysis.

Current work also agreed with findings of Stoner and Ben-Siri (1979), who suggested a different leg action occurs between the inside and outside legs when running a curve, with the outside leg displaying a shorter stride length. Although work by Stoner and Ben-Siri (1979) occurred during the acceleration phase in sprinting, confirmation was gained that a similar adaptation occurred during constant pace curvilinear motion. Adaptation was shown by a decrease in stride length for the outside leg during running ($P < 0.05$) with no corresponding decrease for the inside leg.

In conclusion, the investigation into foot contact time using rearfoot and forefoot footswitches showed no increase in foot contact time with tighter grade of curve during curvilinear jogging or running. When considering altered stride kinematics, a proportional increase in foot contact time was evident, increasing our understanding of increased temporal EMG activity following footstrike. In the outside leg there was an increase in rearfoot contact time as the curve tightened and radius decreased. During running trials, stride length was decreased at the outside leg, but not the inside leg.

Results have shown no significant increase in total foot contact time, and an increased rearfoot contact time in the outside and inside leg. From these results a pattern of proportionally greater foot contact time, with adaptation of ankle motion during curvilinear motion has become evident. A proportionally increased foot contact time was reflected in the greater amount of muscular activity around the stance phase in chapter 3. It was suggested from results in chapter 3 that muscular activity outside of the sagittal plane was used primarily to stabilise the ankle during support. These muscles could be used to alter the path of the centre of gravity during curvilinear motion, yet as curvilinear motion follows a distinctly different path to straight motion, the movement of the segments of the lower extremity must also differ during curvilinear motion. Hamill et al. (1987) presented measurements of rearfoot motion in curvilinear motion, yet provided no information concerning the altered sagittal plane kinematics between linear and non-linear motion. Therefore, more experimental work is required concerning kinematics of the lower extremity during such motion at discrete velocities, and using grades of curve commonly encountered in soccer. To develop a further understanding of the origin of curvilinear performance it is necessary to quantify the movements of various body segments, specifically the lower extremity, and contrast them with those of straight motion. This requires the use of three dimensional kinematics on movements of the lower extremity, and is the focus of the next study.

CHAPTER 5

CHAPTER 5

A Three Dimensional Kinematic Analysis of Straight and Curvilinear Motion

5.1

INTRODUCTION

The results from chapter 4 showed an increased proportional foot contact time to partly explain the increase in muscular activity previously reported around the stance phase in chapter 3. As curvilinear motion follows a different path to straight running, the movement of the segments of the lower extremity must also differ. The alternative path of travel of each limb also suggests asymmetry in curvilinear performance. Such suggestions form the research questions for this study when moving along grades of curve commonly encountered in soccer.

Work relating to kinematic analysis of curvilinear motion by both McMahon and Greene (1979), and Greene (1985) was primarily concerned with maximal sprinting. Assumptions of these studies included the maintenance of constant stride length and stride frequency, which have been shown not to apply to constant paced curvilinear motion by earlier experimental work in this thesis (chapters 3 and 4). Hamill et al. (1987) reported differences in kinematic variables between the inside and outside limbs. Two dimensional kinematics of rearfoot motion were used. Differences were found in pronation and supination angles between straight, inside and outside limb values. A greater mean supination value was displayed in the outside leg when compared to straight motion. Both straight running and the outside leg during curved running showed lower values of pronation than the inside leg, whilst overall the range of pronation/supination movement was found to be greater at the ankle of the outside leg. The calculations of joint displacements in two dimensions are problematic for curvilinear motion. The weakness in reporting values of calcaneal inversion and eversion (rearfoot kinematics) is that pronation and supination occur mainly at the subtalar joint, which is not visible from the rear aspect of the lower leg.

Stoner and Ben-Sira (1979) also suggested different movement patterns between the limbs during curvilinear motion. These findings were verified by data from chapter 4, which showed the outside leg to display reduced stride length whilst running. With a reduced stride length expected to reveal lower angular displacement values, these reductions should be more evident in the outside leg than the inside leg or during straight motion.

To assess the displacement at the joints of the lower extremity during the performance of a stride cycle, and to compare with the angular displacement during straight motion (section 2.4), sagittal plane kinematics must be generated using three dimensional analysis. Such analysis was required due to the constantly changing orientation of the sagittal plane during curvilinear motion. The present study set out to quantify the kinematics of the lower extremity during straight motion and motion at different grades of curvature using three dimensional kinematics. It also investigated whether the adoption of reduced stride length with greater severity of path curvature acted to reduce the angular displacement at the joints of the lower extremity. In addition, angular differences between the two legs during curvilinear motion were to be assessed to quantify the asymmetry between the limbs.

Hypotheses

1)

H_1: There will be a decrease in lower extremity angular displacement values between straight and curvilinear motion.

2)

H_2: There will be a decrease in lower extremity angular displacement values as the grade of curve becomes more severe.

3)

H_3: During curvilinear motion there will be lower angular displacement values at the outside leg, than the inside leg.

5.2
METHOD

Subjects

Eight male soccer players (Age 24.9 ± 4.9 yrs; Mass 77.4 ± 7.9kg) volunteered for the study. Soccer players were targeted due to their familiarity with the movement task. Ethical approval was obtained and the experimental protocol was explained to each subject. Informed consent was obtained and subjects were informed that they were free to withdraw from the study without prejudice at any time. No subjects reported suffering musculoskeletal injuries and all were in good health at the time of testing. Markers were placed over the glenohumeral joint, greater trochanter of the hip, lateral epicondyle of the femur, lateral malleolus, and head of the fifth metatarsal of each leg to aid identification. Each subject required size UK 8 or UK 9 footwear and wore standard soccer footwear (Mizuno Pro-Model) in all trials.

Instrumentation

Environmental conditions remained the same as in previous studies. As results from chapter 3 (section 3.3.2) displayed a greater adaptation at the 5m radius, it was suggested that the increase in curve severity between conditions was not equivalent. Therefore, for the present study, the grades of curve to be investigated were of radius 10 metres, 7.5 metres, and 5 metres, in addition to straight trials. Marker cones placed on the turf surface identified these curvilinear paths. Subject velocity was monitored by infra-red timing lights (Cla-Win Timer, UCC, UK) placed 3m apart at hip height. Subject motion was monitored during each trial using two SVHS video camcorders (Panasonic VHS Supercam AG-DP800E, EG) which sampled at 50 Hz, genlocked with the optical axis positioned at approximately 120 degrees apart. Before each set of trials three dimensional calibration was performed using a 25 point calibration frame (Peak Performance technologies, Englewood, USA). Video tapes were played back using a Panasonic VCR (NV-F75HQ) and a Sharp LCD video projector (XG-3795E) to project the image onto a Terminal display systems high resolution digitising tablet

(TP1067). Image digitisation and analysis was carried out using kine analysis laboratory software (Bartlett and Bowen, 1993) running on an Acorn Archimedes (420/1) computer.

Procedure

Each subject was given ample time to complete a thorough warm up upon arrival at the test location, time to familiarise themselves with the curvilinear paths, and the criterion velocities. At two velocities subjects were required to perform five successful trials at each grade of curve and in a linear path. Target velocities were 4.4m/s for 'jog,' and 5.4 m/s for 'run.' These velocities were determined through investigations reported in section 3.2.1.

The area between the infra-red light gates was calibrated prior to trials at each grade of curve. Subjects were required to pass through the light gates at the criterion velocity, with a trial deemed successful only if a full stride cycle (right foot strike to right foot strike) was performed within the limits of the light gates. The two cameras were event synchronised by a visual signal from the experimenter within the field of view of both cameras.

Data Analysis

From the five trials recorded, the middle three were analysed further. These trials were digitised at 50Hz using a standard 18-point body model, with the two views digitised in sequence. Frame numbers of the key events of heel strike and toe-off were noted. The raw data were smoothed, and derivations of digitised co-ordinates were obtained using a cross-validated quintic spline of all points (Woltring, 1986). This method was chosen over a cross-validated quintic spline of each point on the basis of greatly reduced processing time.

Data extracted at key points of the stride cycle were placed on a spreadsheet (Excel 5.0). Mean values were then computed for each variable, for each subject. To assess differences in movement at each joint of the lower extremity during the stride cycle, mean data were compared statistically using 2 way ANOVA with repeated measures (grade x speed). The ANOVA F test was modified when data lacked sphericity (Coakes and Steed, 1999). Differences were reported at the $P < 0.05$ level.

5.3

RESULTS

In the first instance the angular motion during running is considered qualitatively and then quantitatively. The trends for adaptation of movement as the grade of curve became increasingly severe during running are replicated in the jogging conditions. Where significant differences are noted for grade of curve, changes occurred continuously from straight through 10m, 7.5m, to 5m. Quantitative data relating to jogging are in Appendix E.

The parameters of the stride cycle compared were values at maximum, minimum, heel strike, maximum support and toe-off. An additional parameter of minimum support was noted for the thigh-torso (hip) angle. When considering inside leg values (in this case right leg), two heel strikes were included as the stride cycle was taken from right heel strike to right heel strike.

Running

As the performance of curvilinear motion was considered an autonomous, continuous skill familiar to all subjects, variability between subjects was relatively constant. Joint displacements follow a similar pattern in all subjects except for individual variances in technique probably brought about by differing anthropometric characteristics. Therefore, results presented graphically are representative traces from one subject. Graphical output from jogging trials was similar to running, with results handled in tabular form (Appendix E).

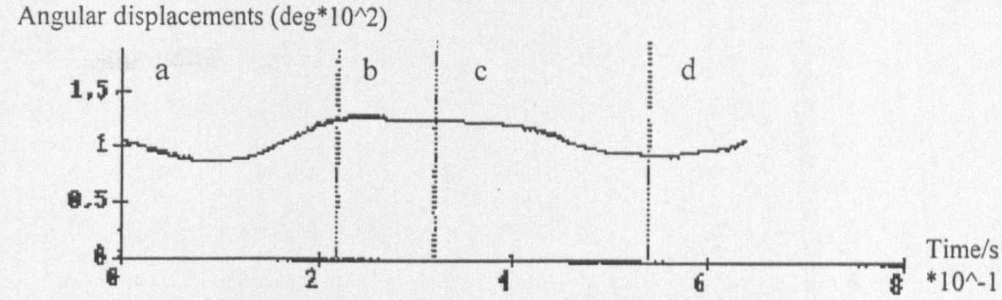

Figure 5.1 Angular displacement of the right ankle (plantar/dorsi flexion) during straight running.
Right heel strike occurs at the y-axis (a). Successive key events marked are toe-off right foot (b); heel strike left foot (c); and toe-off left foot (d).

The graph above depicts the angular displacement of the ankle from the time of heel strike with the right foot. The key events of toe-off for the right foot, heel strike and toe-off for the left foot are marked respectively on the graphical output. Values obtained for angular displacement at the ankle were computed from anatomical landmarks at the knee, ankle and fifth metatarsal head. From heel strike the ankle undergoes dorsiflexion up to mid-stance, where the propulsive phase of the stride cycle begins and plantar flexion ensues. No further displacement occurs until late in the swing phase of the cycle, where further dorsiflexion was noted in preparation for shock attenuation at heel strike.

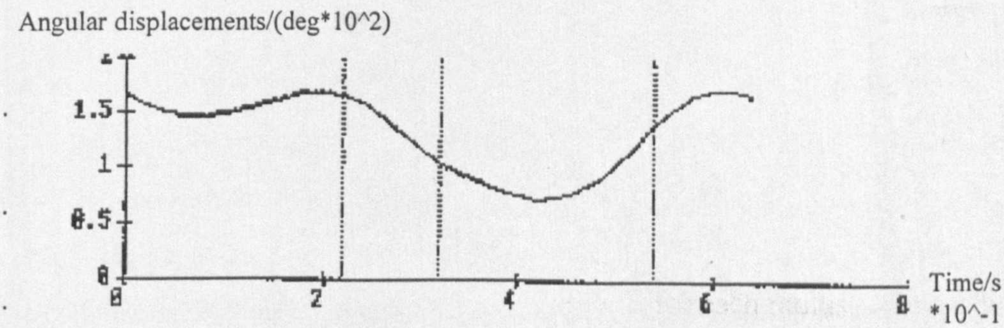

Figure 5.2 Angular displacement of the right knee during straight running. Key events as in figure 5.1.

The angular displacement of the right knee indicates a flexion at the knee towards maximum knee flexion during stance. An extension accompanies the propulsion phase towards toe-off. Once the swing phase is initiated, knee flexion is observed, which served to reduce the inertia of the limb during swing. As the thigh moves forward, the knee is extended in preparation for the impact at heel strike.

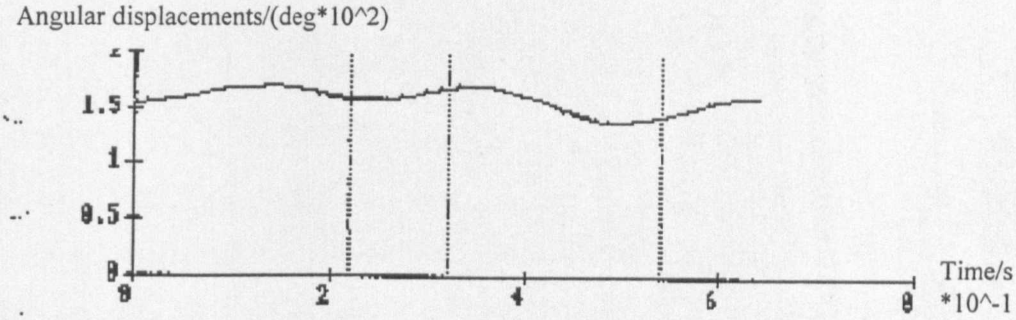

Figure 5.3 Angular displacement of the right hip during straight running. Key events as in figure 5.1.

The angle at the hip was calculated from the displacement between the thigh and the trunk. At heel strike the hip is extending as the thigh moves posteriorly. In some trials an initial phase of flexion is noted as the centre of gravity is decelerated following heel strike. The trial depicted above has no initial flexion phase. During late support the hip begins to flex towards toe-off. After toe-off, the hip once again extended towards the limit of its motion, before beginning to flex and swing forward in front of the torso. As the limits of the forward motion were reached, the thigh began to extend in conjunction with the knee, in a sweeping motion toward heel strike.

Values of angular displacement at the joints of the lower extremity tended to decrease as the grade of curve became tighter, except for at the ankle joint of the inside leg where the only differences observed were with respect to speed. Where significant main effect for grade of curve occurred, decreases in angular displacement were noted for each radius, with adaptation at the 5m curvature greatest. The graphical output at the 5m radius is presented in conjunction with corresponding straight trial for comparison.

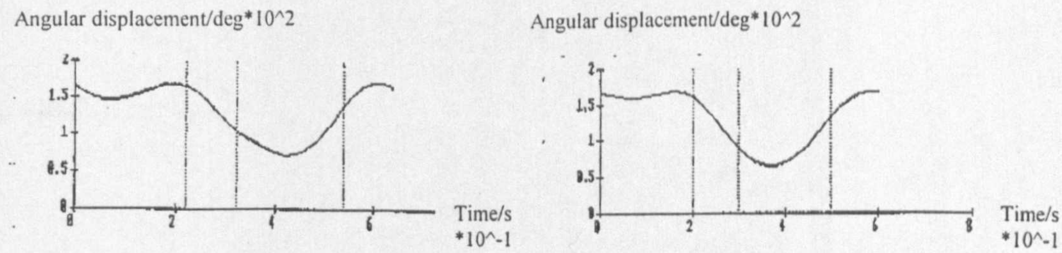

Figure 5.4 Angular displacement at the knee of the inside leg during straight, and 5m curved running. Key events as in figure 5.1.

Several differences were observed between the extremes of grade. Firstly, the overall maximum displacement values were greater in the straight condition. Also noticeable from mean values across all subjects was the greater degree of maximum knee flexion during the 5m curved trial. The trend for maximal values continued at the end of the cycle, as the values at toe-off were also lower at the 5m radius. Temporally, differences were noted between the straight conditions and curvilinear motion, as can be seen from the timing of key events indicated on the graphs. The stride cycle became shorter as the grade of curve increased in severity.

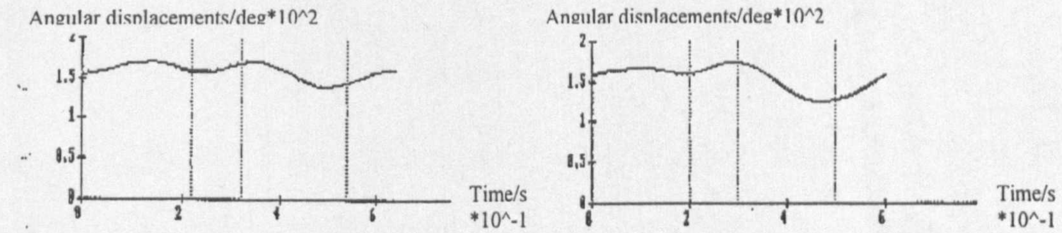

Figure 5.5 Angular displacements at the hip of the inside leg during straight, and 5m curved running. Key events as in figure 5.1.

Mean data for each subject were entered into the analysis for the hip, knee and ankle joints on the inside and outside legs. For the eight subjects during the straight and most severe curvature (5m) mean values for running can be seen in table 5.1, 5.3, and 5.5. Tables 5.2 and 5.4 contain related statistics.

Joint	Trial	Min	Max	Range	1st Heel strike	Min supp.	Max supp.	Toe-off	2nd Heel strike
inside leg hip	st run	131.0	174.7	43.7	150.8	150.0	174.0	163.4	154.8
inside leg hip	5m run	121.5	168.8	47.3	152.2	151.2	167.6	162.4	149.3
outside leg hip	st run	129.2	174.5	45.3	154.5	153.5	172.4	157.7	-
outside leg hip	5m run	128.2	174.7	46.5	153.0	151.1	173.2	164.5	-

Table 5.1 Mean angular displacement (degrees) of the hip during straight and curvilinear motion during running.

Table 5.1 shows that a greater range of hip motion was noted for the inside and outside legs during 5m curved motion. No data are present for heel strike 2 in the outside leg as a stride cycle proceeds from right heel strike to right heel strike, therefore including only one outside (left) heel strike. Statistical differences do exist at the inside leg for values of minimum, maximum, and maximum support. The fundamental shape of the displacement curves remain consistent through straight and curvilinear motion (figure 5.5), although the centre of gravity followed a distinctly different path.

Joint	Value	D.F.	F Value	P Value	Power	Result
Inside leg Hip	Minimum	(3, 21)	4.5	0.014	0.812	st>10>7.5>5
Inside leg Hip	Maximum	(3, 21)	18.22	<0.001	1.00	st>10>7.5>5
Inside leg Hip	Max support	(3, 21)	11.72	<0.001	0.9	st>10>7.5>5

Table 5.2 Statistically significant differences in hip kinematics with increasing curve severity.

Mean data values for the angular displacement of the knee displayed a trend for a decreased range associated with greater flexion as the grade of curve became more severe (Table 5.3). Statistically, values at the knee of the outside and inside legs differed at maximum knee flexion during support (Table 5.4).

Joint	Trial	Min	Max	Range	Heel strike	Max support	Toe-off	Heel strike 2
inside leg knee	st run	58.0	172.1	114.1	159.0	142.5	158.0	167.5
inside leg knee	5m run	57.8	166.1	108.3	161.1	149.8	148.4	160.7
outside leg knee	st run	56.3	169.1	112.8	163.0	144.1	161.3	-
outside leg knee	5m run	60.5	165.9	105.4	155.2	139.7	158.3	-

Table 5.3 Mean angular displacement values at the knee during straight and curvilinear motion during running.

Values at the knee of the inside leg displayed greater adaptation than the outside knee as differences were found for overall maximum, flexion at toe-off and maximum knee flexion during support. This flexion also increased as the grade of curve became more severe. Differences were also noted for the range of motion, with the greater range being noted for the straight conditions.

Joint	Value	D.F.	F Value	P Value	Power	Result
Outside Knee	Max support	(2, 11)	10.54	0.004	0.995	st>10>7.5>5
Inside Knee	Maximum	(2, 16)	3.86	0.037	0.742	st>10>7.5>5
Inside Knee	Max support	(3, 21)	5.7	0.005	0.899	st<10<7.5<5
Inside Knee	Toe-off	(3, 18)	5.04	0.012	0.856	st>10>7.5>5

Table 5.4 Statistically significant differences in knee kinematics with increasing curve severity.

Selected variables showed differences in lower extremity angles with grade of curvature. These variables showed a continuous significant difference as the grade of curve became more severe. Differences were predominantly noticeable in the inside leg of the curve at the hip and knee joints.

Joint	Trial	Min	Max	Range	Heel strike	Max support	Toe-off	Heel strike 2
inside ankle	st run	86.2	134.8	48.6	108.2	86.2	132.3	110.4
inside ankle	5m run	88.3	127.5	39.2	111.4	88.1	121.7	111.4
outside ankle	st run	85.3	139.0	53.7	106.0	85.2	127.3	-
outside ankle	5m run	92.4	134.7	42.3	107.5	92.6	128.3	-

Table 5.5 Mean angular displacement values at the ankle during straight and curvilinear motion during running.

Mean data describing motion at the ankle joint were measured from the displacement between the shank and a line joining the ankle and the fifth metatarsal head. Data were shown to be non-significant for the inside leg, yet differences were noted for the ankle of the outside leg at heel strike, with increasing dorsiflexion as the grade of curve became more severe. The range of motion showed differences, with the straight condition giving greater values. The outside leg ankle angle was smaller at initial heel strike in curved motion. Such a result may suggest differing footstrike characteristics between the inside and outside legs.

Joint	Value	D.F.	F Value	P Value	Power	Result
Outside Ankle	Heel strike 1	(2, 16)	5.28	0.015	0.874	st>10>7.5>5

Table 5.6 Statistically significant differences in ankle kinematics with increasing curve severity.

Data regarding mean kinematic data from jogging trials can be found in tabular form in Appendix E. Statistical treatment revealed differences in the angular displacement values between the two discrete velocities at all of the six joints considered. Overall, the greatest differences were shown for the inside (right) leg of the curve. These differences were evidenced by a greater amount of flexion at the hip and the knee.

Such a result implies less flexion was evident at the outside leg of the curve than the inside leg. The differences showed that joint angles whilst running were lower than whilst jogging, suggesting a greater degree of motion in the running conditions. Differences in displacement with grade of curve were generated from data at both velocities unless specified.

Summarising the results:-

1) There were decreases in angular displacement values between straight and curvilinear motion, therefore the null hypothesis was rejected.

2) There were decreases in angular displacement values as the grade of the curve became more severe, therefore the null hypothesis was rejected.

3) During curvilinear motion there were lower angular displacement values at the outside leg, than the inside leg, therefore the null hypothesis was rejected.

5.4
DISCUSSION

Results showed that running produced significantly lower joint angles at all joints of the lower extremity than jogging, with the exception of the ankle of the inside leg. These results provided a greater range of motion during running, highlighting greater energetic requirements of movement at a higher velocity, in conjunction with results of greater levels of muscular activity presented in sections 3.3.1 and 3.3.2. Values of angular displacement also decreased as a result of curvilinear performance, accentuated as the grade of curve became tighter, specifically at the inside leg. Adaptations at the hip and knee occurred predominantly at the inside leg. When considering the range of motion during the stride cycle, the ankle and the knee showed greater range during straight than curvilinear performance.

When comparing results of the present investigation to those reported on sagittal plane kinematics of straight running at similar velocities, areas of agreement and disagreement were observed. Maximal thigh angle prior to footstrike was reported to be 45° at 4.5 m/s (Cavanagh et al., 1977 cited by Williams et al., 1985). The highest amount of flexion shown in the present study did agree with those findings, as maximal flexion occurred at approximately 132°, which corresponded to an angle of flexion of 48°. However, thigh angles at footstrike do not appear to change appreciably with increased running speed, at least at speeds above 4 m/s (Cavanagh, 1990). At toe-off, thigh angles have been shown to be around -30°, with the thigh continuing to extend several degrees after toe-off. Values in the present study were calculated from the torso to the thigh, whereas in the main, previous work has taken the hip angle as the displacement from the anatomical position. The mean angular displacements obtained at the hip were approximately 163°, which gave a disparity of approximately 48°. In addition to the differences in measurement technique, it would appear that anatomical features may limit thigh extension, since maximal values are near limits for passive range of motion (Luttgens and Wells, 1982). Overall ranges of motion at the hip of 41° tended to agree with previously published work with a mean of approximately 34.4° (Cavanagh, 1990). However, if hip angle had not included the forward leaning torso, mean values would be greater.

Subjects displayed an initial flexion at the hip during early support during selected trials. Previously published literature showed some differences of opinion exist concerning the changes in thigh and hip joint angle during the early phases of support. Miller (1978) showed the thigh segment moving backwards immediately after footstrike, whereas Williams (1980) reported the initial posture is maintained almost to the point of maximum support phase knee flexion. In contrast, Cavanagh (1990) showed the thigh to be moving forward for the first portion of support whilst running at 3.81 m/s, but the pattern disappeared at higher speeds. This would lead to the supposition that the discrepancy is a speed effect, yet data from the present study showed subjects occasionally displaying early support phase flexion. This pattern was not seen in all trials however, and subject velocity was controlled within 5% therefore the pattern would not appear to be affected by speed in the present investigation.

Theoretically one would expect the hip angle to increase throughout the stance phase, yet as the angle was computed from the thigh to the trunk, the observed flexion could be due to movement of the trunk and not the thigh.

The angles created at the knee joint were approximately 15 - 18° at heel strike, which was in agreement with data from Miller (1978) who reported a mean of 19.2°, and Nilsson et al. (1985) with a mean of 14°. Values of maximum flexion were in the range 31° - 38°. Values of maximum knee flexion show a large variation in the literature, yet present data agrees with Nilsson et al. (1985) who reported a mean value of 37°. During the swing phase of the cycle Cavanagh (1990) reported a mean value from the literature of 102.3° for running at approximately 3.81m/s. Accounting for different definitions of the knee angle used in some previous literature, the mean translates to a value of 77.7°. The mean values for the present study were 61° for straight motion, and 63.5° for motion at the 5m radius. Slight differences in the running speed, along with the small sample sizes used in some investigations may account for the discrepancy in results.

The definition of the ankle angle used in the present study differed to those used in previous work. The ankle angle was defined as the angle between the shank and a line connecting the ankle and the fifth metatarsal due to visibility of these points on the recorded image. In contrast, many other studies (Cavanagh, 1990) have defined the ankle angle as an angle between the intersect of the line of the shank and a line connecting the heel and the fifth metatarsal. The values of angular displacement have also been shown not to follow the path of adaptation displayed at other joints during curvilinear performance. The definition of the angle makes comparison with those values present in the literature more difficult. However, altered definition of ankle angle should not be an important factor if changes of angle were compared, yet ranges of motion reported here of approximately 45° during straight jogging still appear distinctly different to the 29° found in the literature (Cavanagh, 1990). These differences may be due to alternative velocities used, or may be attributable to the natural turf surface, which was more compliant than the surfaces previously used during straight overground or treadmill running. Another factor which may cause differing ranges of motion at the ankle was the use of studded soccer footwear. Soccer footwear typically has lower heel height than running footwear used during previous studies. Such an effect is presently unknown as the quantification of the use of this type of footwear has not been addressed in the literature.

Statistically significant differences were found in a number of variables for both factors of speed and grade of curvature. The major finding with increasing speed was the corresponding increase in the range of motion at the joints of the lower extremity. Also, the increase in the relative distance travelled by the segments of the lower extremity through increased ranges of motion will incur a greater angular velocity associated with the greater speed of locomotion. The employment of greater ranges of motion in the lower extremity at higher speed would appear mechanically sound however, as a greater amount of work must be performed to move faster. Therefore, a greater angular change over which to apply force within an individual stride cycle is required. In addition, greater angles of flexion at the knee and hip will reduce moments of inertia about the joints of the lower extremity during gait. Therefore, the greater amount of flexion, the closer the masses of the segments become to the rotating joint. This therefore requires less muscular energy to rotate the lower extremity, enabling more economical locomotion at the faster velocity.

The speed of locomotion was maintained as the grade of curve became more severe, yet maximum knee flexion during support increased at the knee of the outside leg. However, a disparity between the inside and outside legs of the curve again became apparent as results for the knee of the inside leg revealed contrasting findings. Therefore, as the curve severity increased, maximum knee flexion during support increased at the knee of the outside leg, yet decreased at the knee of the inside leg. Increased maximum knee flexion would suggest greater stress being absorbed by the shock attenuating structures of the lower extremity during the support phase. It would be expected that greater stress would be imparted to maintain running at the 5m radius to the same velocity as running straight. Therefore the increased maximal knee flexion in the outside leg is unsurprising. However, the difference between the inside and outside legs would indicate differing functions of the two limbs during curvilinear performance. The overall pattern at the hip and knee was for greater flexion at the inside leg throughout the stride cycle. Such data can be explained by the need for foot clearance from the ground during the swing phase, combined with the effect of the 'lean' associated with curvilinear motion. Increased flexion at the outside leg may therefore indicate greater force bearing by that limb during curvilinear motion. Decreased knee flexion at the inside leg could be a function of different footstrike characteristics occurring between limbs.

The effect of increasing the speed of locomotion brought about significant differences at all joints except the ankle of the inside leg. Results showed that motion at the ankle joint was relatively constant during curvilinear motion. However, preliminary evidence suggested an increased tendency for forefoot-strike occurs with increasing curvature. Such a footstrike would cause the metatarsal-phalangeal joint rather than the ankle to act as a fulcrum for rotation when a foot-flat position is achieved during support. This may mean that the foot uses a different axis during the propulsion phase between the inside and the outside limb of the curve. The absence of adaptation in the kinematics of this joint implies that range of motion was not increased as at the other joints of the lower extremity. However, as a decrease in the angular displacement has become apparent for curvilinear performance in general, the implication of no significant difference at the ankle of the inside leg could be an important feature to the performance of this type of movement.

Another key finding at the ankle occurred at the outside leg. All significant differences with respect to speed showed lower angular displacement in the running conditions, except for the ankle at the outside leg. Findings at this joint showed greater angular displacements at key events during running. These results imply greater plantar flexion during running than jogging for the outside ankle. When linked with the findings for the ankle at the inside leg, indications are that the ankle joint is a key site where movement differs from the general pattern of the lower extremity during curvilinear motion.

The angular displacements noted at the ankle joints were measured as the angular change between the segments of the shank and the ankle-toe line. Therefore, it may be that altered angular displacements in this plane occur to facilitate movement at the ankle in another direction. Qualitative observation of increased ab/adduction at the ankle was noted during curvilinear performance, as was lateral rotation of the femur. These additional movements are likely to be critical in the performance of curvilinear motion to enable the transmission of force to the ground in the correct direction. Information regarding the orientation of the lower extremity during the support phase would yield further understanding of the footstrike characteristics of the inside and outside limbs during curvilinear motion. However, the measurement techniques employed did not enable the computation of such variables.

5.5
CONCLUSION

Curvilinear motion on a natural turf surface was performed to gain an insight into the kinematics of movement at two discrete soccer match velocity conditions. It was thought that altered temporal muscle activity patterns (shown in chapter 3) would lead to different movement patterns of the segments of the lower extremity, and provided the research question for the present study. Differences could be identified in altered kinematics. Different movement patterns between the limbs during curved running had also been suggested in the literature (Stoner and Ben-Sira, 1979). Therefore, the differences in kinematics between straight motion and motion at three grades of curvature was investigated for both inside and outside legs. Results were presented for straight and curvilinear motion at the 5m curvature, with graphical output described for a typical subject.

Differences were shown in the angular displacement values. For maximal, minimal, and values at key events during the stride cycle, angular displacement values at the lower extremities were greater when jogging than running, hence showing a greater range of motion at the joint for the running velocity. As the grade of curve was altered, differences in selected values of angular displacement at the lower extremities were seen. Most differences occurred for the inside leg of the curve, at the hip and the knee, with values tending to reduce (ie. greater flexion) with curve severity. However, no differences were noted for the ankle of the inside leg. It was surprising that no altered adaptation took place at the inside ankle as simple inspection of an athlete running in a curved path suggests a more prevalent forefoot strike in that limb. Perhaps this joint may not alter its range of movement during curvilinear motion, yet may be a joint where key adaptation occurs outside of the sagittal plane during motion of this type. Kinematic details of movement outside of the sagittal plane would be required if these assumptions were to be tested.

Overall, values obtained for straight motion were in agreement with the literature on straight treadmill and overground running, when comparable angle definitions were used. When comparing the two extremes of motion - straight and 5m curve, the range of motion at each joint of the lower extremity tended to decrease in the curvilinear condition. This occurred due to the shorter stride length as curve severity increases, which was evident from earlier experiments (chapter 3 and 4). Also accompanying the decrease in stride length was an increase in stride frequency. These factors ensure less distance was covered in each step during curvilinear motion compared to straight running. The overall body movement was therefore smaller, with an associated reduction in range of motion at the joints of the lower extremity.

Adaptation of body segment movement during curvilinear performance has been identified in the lower extremity. Differences have been noted between the actions of the two limbs during curvilinear performance, and their possible effect on the mechanisms of curvilinear progression has been speculated upon. To understand the relative contribution of each leg to the creation of the centripetal force required for curvilinear motion, it is necessary to measure ground reaction force. In addition, accurate measures of foot contact time may be obtained from these measurements to further enhance knowledge of the mechanisms involved in curvilinear motion.

CHAPTER 6

CHAPTER 6

An Investigation Into the Kinetics of the Ground Reaction Force During Straight and Curvilinear Motion on a Natural Turf Surface

6.1

INTRODUCTION

The differences noted from the electromyographical signals in chapter 3, in addition to the kinematic measures of chapters 4 & 5, showed differences between the inside and outside legs of the curve during curvilinear performance. Qualitative observation would support the quantitative data by suggesting a major difference between the two limbs of the lower extremity occurs at the ankle joint. Although general body rotation must occur to maintain a sagittal plane coincident with the direction of movement, the continuous alteration of the body position is transmitted through the ankle joint complex to the ground. Results from chapter 5 suggested an alternative function of the inside and outside limbs.

The contribution of target muscles from chapter 3 showed no difference in the magnitude of muscular contraction during straight and curvilinear motion. These results would suggest that no difference in applied muscular force occurs during curvilinear motion. However, temporal activity values were altered and served to provide a greater duration of activity during curvilinear motion. These adaptations would not suggest a greater peak muscular force, but may suggest a greater total muscular force involvement in curvilinear force production. In addition, only selected superficial muscles were measured at the lower extremity, and therefore may not represent the holistic pattern of activity. It would be expected that the total muscular force would be an indicator of the contributions to curvilinear performance by the two limbs. Total force values are a summation of the force measured in the three orthogonal axes and have historically been measured by force plate analysis. Comparison of these values would enable analysis of the contribution made by each limb to curvilinear motion and formed a research question for the present study.

Previous work conducted on force plate analysis during gait is considerable, and key investigations are summarised in section 2.6. The analysis of forces associated with curvilinear performance, and more specifically curvilinear performance in soccer play has received little attention to date. As long ago as 1987, Hamill et al. stated that curved path locomotion may subject individuals to unique stresses and that research examining this type of movement has been largely neglected. The main reason for this would appear to be the methodological difficulties associated with the acquisition of ground reaction forces in curvilinear motion. Many studies investigating the area of soccer have not reported the surface used, but work by Saggini et al. (1992) and Saggini and Vecchiet (1994) have investigated the ground reaction forces during straight running of soccer players on natural turf. Other investigations concerning force analysis in soccer (Rodano et al., 1988) have focused on the soccer kick, but have not used a natural turf surface or studded soccer footwear.

The only previous study found which quantified ground reaction forces during curvilinear motion was Hamill et al.'s (1987) which addressed the ground reaction forces during negotiation of athletics bend running at 31.5 metres radius. Whilst typical curvilinear motion in soccer differs due to increased curve severity and surface used, these values do provide a baseline data set. Results displayed differences in all vertical ground reaction force variables describing the impact phase, with the outside leg always displaying the greater values. Differences were also noted in all mediolateral components, with values in curved running always greater. However, measurements were made using one force platform only, therefore details of successive footfalls of the inside and outside leg of the curve could not be obtained.

Curvilinear motion could be considered as a series of crossover and side-step cutting manoeuvres, which can often be seen in soccer play. Following investigation into 45° and 90° cutting movements, Schot et al. (1995) concluded that two mechanisms may be used to generate these movements, the theory to which were covered in chapter 1. Andrews et al. (1977, cited by Schot et al., 1995) suggested such actions were accomplished mainly through torque generated by the torso, pelvis and lower extremity musculature and applied to the ground. In contrast, Hamill et al. (1987)

claimed that the change of direction in curvilinear motion on a track was predominantly created by increased mediolateral force. Conclusions suggested the increase in mediolateral force during curvilinear motion was the principal mechanism.

Whilst measurements of the energetics at the shoe-surface interface would provide important information to the understanding of curvilinear motion, soccer provides an additional variable of the specialist footwear used. Whilst it is widely believed that studded soccer footwear is worn to increase friction at the shoe-surface interface, enabling greater acceleration and quicker movement over a natural turf surface (Torg et al., 1974), there remains no evidence of the mechanisms for altered movement. If greater friction were obtained at the shoe-surface interface, it would be expected that greater forces could be imparted whilst wearing studded soccer footwear. The quantification of ground reaction forces whilst performing movements representative of soccer match conditions would hopefully elicit such information and enable further understanding of the interaction at the shoe-surface interface in soccer.

The use of force plate analysis would provide important data regarding times of support and non-support, in addition to an understanding of the energetics involved at the inside and outside legs during curvilinear motion. The ground reaction force variables that were considered were those suggested by Bates et al. (1983). Key variables of interest were vertical ground reaction force, as an indication of impact characteristics between straight and curvilinear motion; mediolateral forces to assess the contribution of each limb to the centripetal force; total ground reaction force to gain an insight into the energetic transfer required for curvilinear compared to straight motion; foot contact and ballistic air time to enable comparison of results with earlier studies (chapters 3 and 4). The experimental work in this chapter was to quantify ground reaction forces in successive footfalls during straight and curvilinear motion on a natural turf surface, with the aim of assessing the relative contribution of each limb to curvilinear motion. The investigation would look at the forces occurring when soccer players wear indoor soccer footwear and standard soccer footwear to assess the difference of the altered shoe-surface interface conditions on the ground reactions force during straight and curvilinear motion.

Hypotheses

1)

H_1: Total ground reaction force values will be greater at the outside leg of the curve.

2)

H_2: Total ground reaction force values will increase from straight to curvilinear motion.

3)

H_3: Total ground reaction force over two consecutive footstrikes will be greater during curvilinear motion than straight motion.

4)

H_4: Mediolateral ground reaction forces will increase from straight to curvilinear motion.

5)

H_5: Mediolateral ground reaction forces will be greater at the outside leg than the inside leg.

6)

H_6: Foot contact time at the outside leg will increase from straight to curvilinear motion.

7)

H_7: Foot contact time at the inside leg will increase from straight to curvilinear motion.

6.2
METHOD

Introduction

In the previous studies the greatest differences were reported between the two extremes of curvilinear grade. Therefore, this study investigated only straight motion and curvilinear motion at a 5m radius. To enable comparison of data with those collected earlier, it was necessary to collect electromyographical data to ensure temporal muscular activity was comparable. Data were collected from the medial and lateral heads of the gastrocnemius because the medial head showed greatest adaptation, whilst the lateral head provided backup data.
For comparison with data from earlier experiments in the thesis, it was necessary to show that a similar type of movement was taking place. If temporal data concerning the activation and cessation of muscle activity during the stride cycle were similar, findings from earlier work could be compared to the present investigation, and *vice versa*.

Subjects

Six male soccer players (age 25 ± 4.73 years; mass 79.7 ± 7.17 kg) volunteered for the study. Subject numbers were lower than in previous chapters due to the volume of data to be collected. Soccer players were preferred for their familiarity with the patterns of curvilinear motion required. Ethical approval was gained, experimental protocol was explained to each subject, and informed consent obtained. Subjects were reminded that they could cease participation in the study at any stage without prejudice. All subjects had footsize UK 8 or UK 9 and were provided with standard six-stud soccer boots (Mizuno Pro Model). Subjects also provided their own indoor soccer shoes. No subjects reported any musculoskeletal injuries at the time of testing.

Instrumentation

Force Platform Rig Dimensions and Construction

Movements in soccer occur linearly or non-linearly. As a consequence of this, a force platform rig must enable the measurement of ground reaction forces during both forms of movement. The gait of a soccer player will vary depending on the anthropometrics, preferred stride length, speed of locomotion, and direction of travel of the athlete. To enable measurement, and thus give representative results, any rig construction must have the capacity to accommodate typical soccer players. No literature exists regarding the ground reaction force at successive footfalls on a natural turf surface. Therefore it was necessary to construct a rig which could contain a minimum of two force platforms to measure consecutive footfalls in straight and non-linear motion.

With respect to linear motion, computed values of stride length were taken from chapters 3 and 4, in addition to values obtained from the running literature. The step length was assumed to be approximately 50% of these values. The overall length of the rig needed to be sufficient to accommodate the extremes of step length values. For considerations of stride length, the length of the rig must be a multiple of the force platform dimensions. The dimensions of the force platforms to be used (Kistler type 9281B) were 600mm length x 400mm width. From the data it was decided that 3m (5 x force platform lengths) was sufficient for the considered population.

To measure non-linear in addition to linear motion, the position of a force platform outside of the original 3m line was required. From notational analysis conducted in chapter 3, it became evident that apart from tackling and turning, a 5m radius of curve was approximately the most severe encountered by soccer players. Therefore, placement of a second force plate ought to enable measurement of consecutive footfalls in curvilinear motion up to a severity of 5m radius. It was calculated that to enable this, plate position must be allowed up to two plate widths from the original plate positions. The force platform rig was to be used for a number of subjects with repeated trials. Such repetition would cause a large amount of wear on the natural turf around the rig. As straight motion could be performed from either direction along the set of 5 force platform positions, it was decided that to prevent excessive wear on the surrounding turf, the possibility of performing curvilinear motion at either end of the rig should also be permitted.

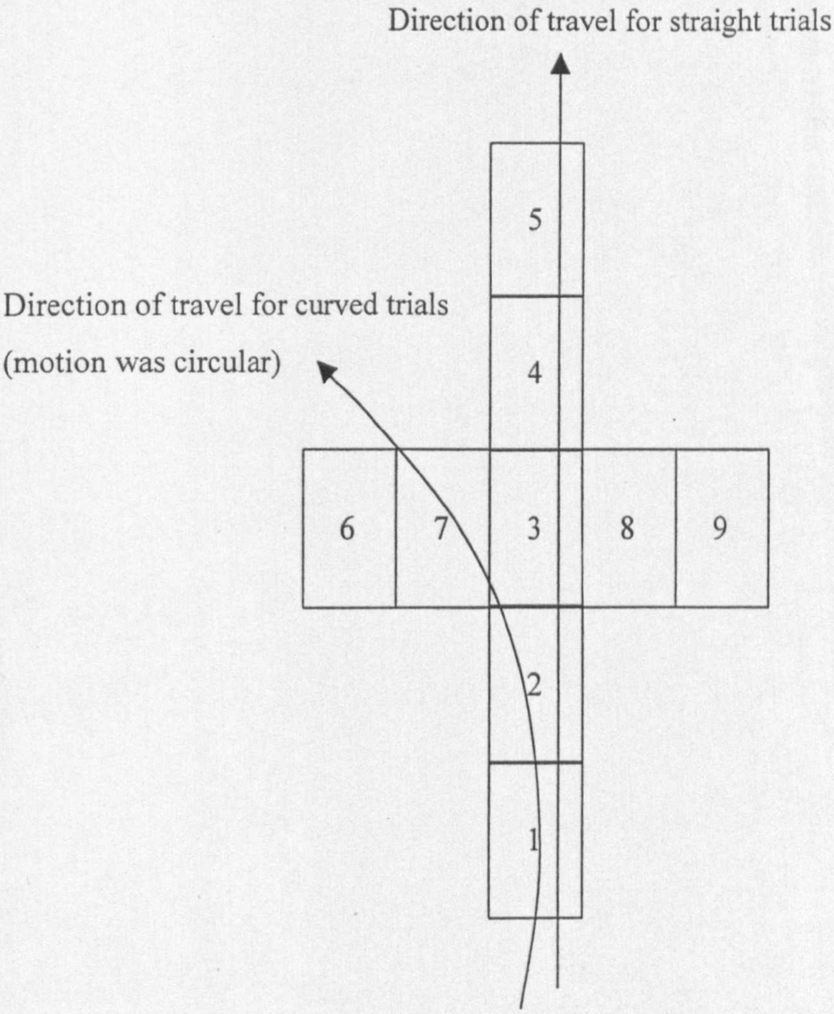

Figure 6.1 Layout of possible plate positions in the force platform rig.

The inclusion of positions 8 and 9 enabled surrounding turf wear to be minimised by enabling curvilinear motion to take place at either end of the rig with one plate in position 1 or 5, and another in position 6 or 9.

In addition to those positions shown, it was realised that some subjects may have possessed a step length which would mean that a second plate positioned in either of two adjacent positions would consistently cause the footstrike to miss the platform by half of a footlength. Therefore, the rig was constructed (University College Chichester) to allow not only the positioning of plates as shown, but also every half plate length or half plate width. Hence, measurement of consecutive footfalls of all subjects could be undertaken.

All testing was performed on a regularly cut natural turf surface in an area of approximately 40 x 40m surrounding the force platform rig to replicate soccer pitch conditions. Turf moisture was assessed prior to each testing session using a soil wetness meter (Rapitest, UK), with testing only proceeding if a minimal reading of 3 was obtained (range 1 - 4) to allow stud penetration. If soil moisture was too low, the surrounding area was irrigated until a value of 3 was attained.

Recording of ground reaction forces was achieved using 2 Kistler piezo-electric force platforms (type 9851B) mounted with the force platform rig in a natural turf surface. The platforms were connected via cables to KIAG Swiss electronic units for multicomponent force measurement, analogue/digital converters and then to a personal computer (Viglen 4DX266) running GmbH MYO-DAT 5.0 software for Kistler force platforms. The force platforms were covered with natural turf plates which were bolted to the force platforms.

Kinematics

The circular path of radius 5m was identified on the turf with marker cones and subject velocity was monitored using infra-red light timing gates (Cla-Win timer, University College Chichester, UK) placed 3m apart at hip height. Subject lower extremity motion was monitored using two SVHS video cameras (Panasonic VHS Supercam AG-DP800E, EG), genlocked at 50Hz, with the optical axis positioned approximately 120 degrees apart. Three dimensional calibration was performed using a 17 point calibration frame (Peak Performance technologies, Englewood, USA) placed between the two force platforms. The position of the foot during contact was recorded along with an audio record of the experimenter's comments using an additional video recorder (Sony Hi-8). Digitisation took place using a Panasonic VCR (NV-F75HQ) through a Sharp LCD video projector (XG-3795E), onto a Terminal Display Systems digitising tablet. Data were digitised and analysed using an Acorn

Archimedes (420/1) computer running Kine analysis laboratory software (Bartlett and Bowen, 1993).

EMG

Muscular activity was recorded at the medial and lateral heads of the gastrocnemius of the right leg. Electromyography equipment was as reported in chapter 3.

Procedure - Setup

Prior to the testing sessions, the natural turf covering to the force platform surface was prepared. Approximately ten turf force platform plates were prepared a minimum of twelve hours prior to a testing session. If some turf plates were not used in a testing session they were maintained and irrigated until a subsequent session with all turf plates maintained at a minimum soil wetness value 3 using the soil wetness meter (Rapitest, UK). As some subjects would require plate mounting between two rig positions it was necessary to prepare some turf plates with half the dimensions of the force platform. Four such plates were prepared for each session.

Two force platforms were mounted in the rig and the mounting verified with a spirit level. The first in position 1, and the second dependent on the stride characteristics of the subject. The force platform cables were then attached to the force platform amplifiers (Kistler, Alton, UK) and in turn to the personal computer. Amplifiers were given approximately 30 minutes to reach operating temperature and their stability then checked. The remaining positions on the rig were then covered using metal fillers, each of which was engineered to half the dimensions of a force platform. When all fillers were in place and the amplifiers were ready for use, the turf plates were bolted to the surface of the platforms and the fillers. The mean overall depth of the turf plates was approximately 32mm, and this value was entered to the Kistler amplifiers as the Z offset.

During the performance of curvilinear motion it was expected that subject footstrike would not occur in line with the y-axis of the force platform. Therefore, measurement of the angle of the foot in relation to this axis was calibrated prior to experimentation. A metre rule and a goniometer were used to quantify the deviation from the axis in 10 degree intervals. When the video recording of the trials was replayed, an acetate of the calibration was overlaid onto the screen to enable quantification of the foot angle during contact.

To calibrate the movement space above the force platforms for three dimensional analysis of the lower extremities, the 17 point calibration frame must be viewed by both cameras. To aid in identification of each of the white spherical points, a black plastic sheet was placed behind the frame in the view of each camera to aid contrast.

Electrode application for the muscles of the gastrocnemius medial and lateral heads of the right leg was accomplished according to the procedure outlined in chapter 3, with a reference electrode placed on the patella. Impedance values were recorded for each muscle, with electrodes re-applied if values exceeded 10kΩ. A synchronisation switch was connected to the computers controlling both EMG and force platform systems to enable sampling of both computers to be synchronised. Temporal values of heel strike from the force platforms could then be applied to the EMG data.

Data Acquisition

Each subject was given adequate time to warm-up, stretch, and familiarise themselves with the target velocities required. During the familiarisation procedure, the experimenter noted footstrike positions and, if necessary, adjusted the position of the second force platform to coincide with the subject's step length. Five successful trials of consecutive foot contacts at the target velocities of 'jog' at 4.4m/s ± 5% and 'run' at 5.4m/s ± 5% were required in both straight and curvilinear conditions. Subjects performed curvilinear trials in an anticlockwise direction and collected data for inside and outside limbs. Subjects were instructed to maintain a smooth running pattern at the designated velocity and to look ahead so as not to target the platforms. The starting footwear condition was randomised for each subject, as was the order of straight and curvilinear trials. Trials were deemed unacceptable if the target velocity was not reached, the subject did not make full contact with both platforms, or the experimenter deemed the subject executed an abnormal stride by targeting the platform. Testing continued using a set of turf plates until they became damaged or exhibited wear to the turf surface. When this occurred testing was halted whilst the turf plates were replaced. Turf plates typically lasted 25 trials before replacement was required.

Data analysis

Force platform data were normalised for body weight to allow for comparison within and between subjects. Three of the successful five trials were analysed, the first and last were omitted from the analysis. From the raw data, 20 key measures were extracted for each footstrike as indicated in Table 6.4. The values were compared between straight motion and the inside and outside leg, in curvilinear motion. Mean values for each subject were entered into a one-way ANOVA statistical model with repeated measures.

Three dimensional kinematic data were digitised at 50Hz with a user defined 10-point model, with the two views digitised in sequence. The points digitised were hip, knee, ankle, heel and distal end of big toe for both lower extremities. Frame numbers of the key events of heel strike and toe-off were noted. The raw data were smoothed, and derivations of digitised co-ordinates were obtained using a cross-validated quintic spline of all points. The angle of the shank with respect to the vertical in the saggital plane was computed and was taken as the shank angle at footstrike. Displacement co-ordinate values were taken from the graphical output, which following data smoothing displayed a deflection point which represented footstrike. These data were computed and placed on a spreadsheet (Excel 5.0). Mean values were then computed for shank angle, for each subject. To assess differences in shank angle of the lower extremity at footstrike, mean data were compared statistically using 2-way ANOVA with repeated measures (grade x speed). The ANOVA F test was modified when data lacked sphericity (Coakes and Steed, 1999). Differences were reported at the $P < 0.05$ level.

For analysis of electromyographical data, temporal values were of interest. Footstrike times were extracted from the synchronised force platform data and entered into the

EMG analysis to indicate the start of the stride cycle. A typical trace was analysed from each condition, according to the criteria that there was little data noise within the raw EMG, and muscle activity displayed a representative pattern. Cessation of activity was identified manually for both the gastrocnemius medial and lateral heads. Values for cessation of activity following footstrike were normalised to a percentage of the stride cycle and entered into a spreadsheet format (Microsoft Excel 5.0). EMG data were then used for verification of the similarity of actions between the present study and chapter 3 using independent samples t-tests.

6.3
RESULTS

Force-time curves from the data collected showed minor intra-subject variation. Traces for two consecutive footstrikes of a 703N subject during straight jogging in soccer boots are depicted below.

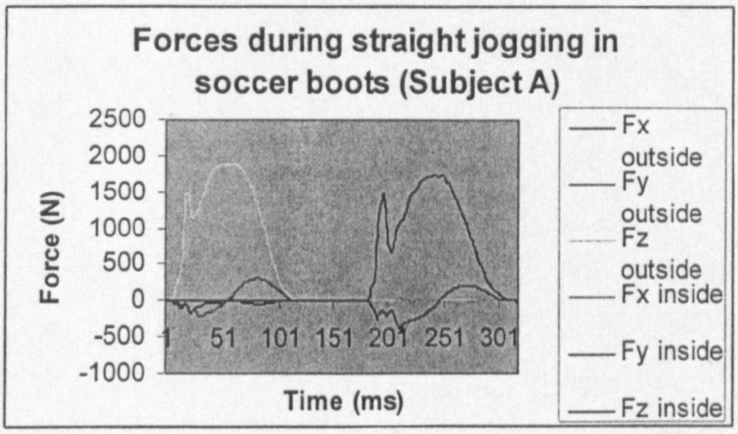

Figure 6.2 Force-time curves (self-scaled) of a typical trial of straight jogging in soccer boots.

Channel 1: Fx outside foot contact

Channel 2: Fy outside foot contact

Channel 3: Fz outside foot contact

Channel 4: Fx inside foot contact

Channel 5: Fy inside foot contact

Channel 6: Fz inside foot contact

The vertical ground reaction force trace (Fz) displays a characteristic impact peak (mean 2.5 *BW* footstrike 1, mean 2.6 *BW* footstrike 2), followed by a minimum corresponding to knee flexion, and an active peak to correspond with propulsion (mean 2.9 *BW* footstrike 1, mean 2.8 *BW* footstrike 2). Both footstrikes in the straight condition displayed a predominantly medial force. The anterior-posterior (Fy) component displayed variability during the braking phase (mean 0.59 *BW* footstrike 1 and 2). Propulsive forces were slightly lower than braking (mean 0.41 *BW* footstrike 1, mean 0.37 *BW* footstrike 2).

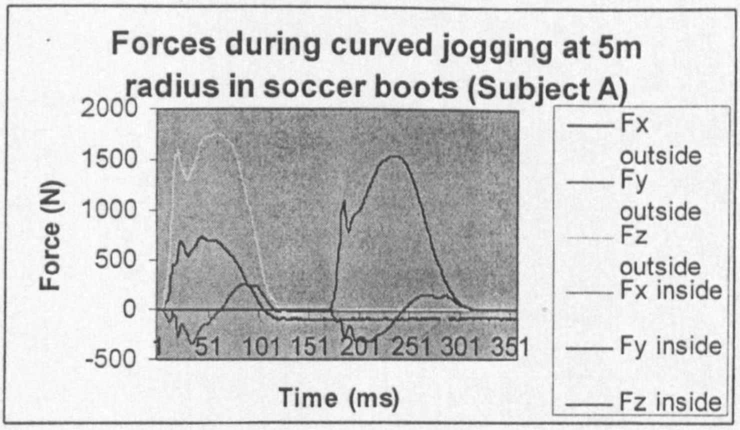

Figure 6.3 Force-time curves (self-scaled) of a typical trial at 5m radius jogging in soccer boots with 1st footstrike by outside leg and second footstrike by inside leg.

Channel 1: Fx outside foot contact

Channel 2: Fy outside foot contact

Channel 3: Fz outside foot contact

Channel 4: Fx inside foot contact

Channel 5: Fy inside foot contact

Channel 6: Fz inside foot contact

For curvilinear jogging at the 5m radius, vertical and anterior-posterior traces displayed similar patterns to straight motion. Vertical traces showed impact peaks values of 2.0 *BW* and 1.7 *BW* for outside and inside limbs respectively. Active peak values were 2.5 *BW* and 2.3 *BW* for outside and inside legs respectively. These values were considerably lower than during straight motion, especially for the impact peak. Again fluctuations were evident during the braking phase of footstrike in the anterior-posterior traces, as in the straight trials.

The greatest differences were noted in the mediolateral direction, primarily with the outside leg ground reaction forces generating acceleration toward the inside of the curve. In comparison to straight mediolateral traces as depicted in figure 6.2, curvilinear mediolateral traces were similar in shape to the vertical ground reaction force and across all subjects reached mean maxima of 1.0 *BW* and 0.65 *BW* for outside and inside limbs respectively. These values greatly exceed those for straight motion.

Statistical analysis was performed using a series of one-way ANOVA techniques. The results showed differences existed in a large number of the 20 considered parameters adapted from Bates et al. (1983). Comparisons were made between values of the straight, and outside and inside leg footstrikes during curvilinear motion. Care must be given to interpretation where minimum values are considered, as differences between legs where one value is lower can infer a greater force applied, yet in the negative direction. Where differences exist, the direction is highlighted. With the exception of the moment about the vertical axis (highlighted in grey), outside leg values always exceeded inside leg values. However, for some parameters the inside leg value was significantly less than both the straight and outside values, whereas for other parameters the outside leg value was significantly greater than both the straight and inside values. Where differences between all three footstrikes occur, the order of magnitude is given.

Overall total force maxima and average values were less at the inside leg, than the outside leg or during straight motion. There were no significant differences according to footwear, whether boots or shoes. In curvilinear motion generally all significant differences in table 6.4 related to lower values being recorded for the inside leg, probably reflecting the lower muscular forces generated at the inside leg than the outside leg. This latter finding confirms EMG data reported in chapter 3, which concluded that the predominant muscular adaptation to curvilinear motion occurred in the outside leg.

	BOOTS		SHOES	
	JOG	RUN	JOG	RUN
Time to 1st max Fz	☆out larger		☆out larger	
1st maximum Fz	☆in less	☆in/out less	☆in less	☆in less
Time to 1st minimum Fz	☆in less	☆in less	☆out larger	
1st minimum Fz			☆in less	☆in less
2nd maximum Fz	☆in less	☆in less	☆in less	☆in out st
Fz Impulse		☆in less		☆in less
Mz maximum		☆out larger		☆out larger
Mz minimum	☆in less	☆in/out less	☆st larger	
Average force		☆in/out less	☆in less	☆in out st
Total Force max	☆in less	☆in less	☆in less	☆in less
Total Force Ave		☆in less	☆in less	☆in less

Table 6.4 Statistically significant differences during straight and curvilinear motion in vertical and total ground reaction force parameters. (☆ represents statistical difference at $P < 0.05$ level).

Note; st = straight; in = inside leg; out = outside leg. Fz impulse = vertical impulse (Ns); Mz = moment about vertical axis (Nm); Total force = $\sqrt{}Fz^2 + Fy^2 + Fx^2$.

In curvilinear running the Fz impulse was less for the inside leg indicating the greater muscular forces generated within the outside leg and related greater inward turning moment about the vertical (Mz) for the outside leg. The inside and outside legs (ie. curvilinear motion) showed differences in minimum Mz. These differences correspond to larger amounts of outward rotation in the outside leg and inward rotation at the inside leg than during straight motion.

During curvilinear jogging the time to first maximum was notably longer for the outside leg, possibly indicating that in curvilinear motion that the loading rate of the inside leg and that occurs in straight motion is greater than for the outside leg.

	BOOTS		SHOES	
	JOG	RUN	JOG	RUN
Total Force Max straight	5.91 ± 0.25	5.87 ± 0.23	5.79 ± 0.24	5.75 ± 0.21
Total Force Max curve	5.13 ± 0.19	5.07 ± 0.30	5.19 ± 0.19	4.89 ± 0.21
Total Force Ave straight	3.38 ± 0.11	3.53 ± 0.12	3.41 ± 0.14	3.49 ± 0.10
Total Force Ave curve	3.08 ± 0.16	3.08 ± 0.15	3.03 ± 0.12	3.02 ± 0.13

Table 6.5 Summed Total Ground Reaction force values over two consecutive footfalls during straight and curvilinear motion at a 5m radius.

To consider the total force over a complete stride cycle, the total force from two successive footstrikes was summed. Results from statistical analysis of these summed data showed significantly greater maximum total force in the straight condition compared to motion at the 5m radius. Average total force values were significantly greater during straight running, but not in jogging. No footwear differences were observed.

	BOOTS		SHOES	
	JOG	RUN	JOG	RUN
Fx maximum	☆out>in>st	☆out>in>st	☆out>in>st	☆out>in>st
Fx minimum	☆st less	☆ st less	☆ st less	☆ st less
Fx Ave	☆out>in>st	☆out>in>st	☆out>in>st	☆out>in>st
Fx Impulse	☆out>in>st	☆out>in>st	☆out>in>st	☆out>in>st

Table 6.6 Statistically significant differences in mediolateral ground reaction force variables. (☆ represents statistical difference at $P < 0.05$ level).

For motion in a curve, ground reaction forces accelerating the body toward the centre of the curve will be directed towards the medial side of the outside foot, and the lateral side of the inside foot. These forces would be represented as positive Fx values for both the outside foot and the inside foot. Generally in both jogging and running in curvilinear motion, forces were greater for the outside leg than the inside leg, and both exhibited mediolateral forces greater than those occurring in straight motion.

	BOOTS		SHOES	
	JOG	RUN	JOG	RUN
Fy minimum		☆ in/st less		☆st less
Braking Impulse		☆out larger		☆ out larger
Time to 0 crossing				
Fy maximum	☆in less	☆in less	☆in less	☆in less
Propulsion impulse	☆in less	☆in less	☆in less	☆in less

Table 6.7 Statistically significant differences in anterior-posterior ground reaction force variables. (☆ represents statistical difference at $P < 0.05$ level).

Time to the crossing of the x-axis for anterior-posterior force did not differ between conditions and remained at approximately 50%. Maximum propulsive force and propulsive impulse showed the outside legs and straight conditions displayed greater values than the inside leg and the effect was not footwear related. In running, mean braking forces were greater in shoes for straight motion (0.63 $BW \pm 0.09$ S.E.) than for the inside leg (0.44 $BW \pm 0.06$) and outside leg (0.43 $BW \pm 0.04$) in curvilinear motion. The mean braking impulse during running was greater on the straight and for the inside leg on the curve than for the outside leg on the curve. Possibly this was related to the shank angle at footstrike (Table 6.9) which was nearer vertical at the outside leg of the curve. Data from chapter 4 also showed that the outside leg displayed a significantly ($P < 0.05$) shorter stride length at 5m radius, whilst changes at the inside leg remained non-significant.

Condition	Outside leg (degrees)	S.E.	Inside leg (degrees)	S.E.
Jog Shoes	1.11	1.11	27.5	3.70
Jog Boots	1.67	1.52	29.67	2.86
Run Shoes	6.11	3.03	29.17	4.12
Run Boots	6.67	3.19	34.0	2.18

Table 6.8 Mean values of heel-toe angle with respect to the anterior-posterior axis of the force platform during curvilinear motion at 5m radius.

The table above displays foot abduction angles with respect to the anterior-posterior axis of the platform. As shown by the mean data for foot contact angle, the inside foot abducts further from the anterior-posterior axis of the platform than the outside foot.

Half of the subjects displayed no abduction at all in the outside leg of the curve. However, it must be remembered that the experimental setup meant that the first platform was situated with the A-P axis tangential to the curve, whereas the second platform created an angle of approximately 20° for each subject with the tangent of the curve. However, when the platform orientation is considered the abduction angle at the inside foot remains larger than the outside foot counterpart.

	Inside leg angle (degrees)	Outside leg angle (degrees)
Shoes straight jog	71.3 ± 1.51	77.4 ± 1.08
Shoes 5m jog	60.4 ± 1.95	65.5 ± 0.95
Shoes straight run	72.4 ± 1.70	75.1 ± 0.91
Shoes 5m run	53.4 ± 2.76	56.6 ± 3.37
Boots straight jog	75.4 ± 1.49	79.8 ± 1.27
Boots 5m jog	60.3 ± 1.02	66.0 ± 1.0
Boots straight run	75.0 ± 1.87	76.3 ± 0.63
Boots 5m run	53.8 ± 2.15	60.2 ± 1.44

Table 6.9 Table showing mean shank angle (± S.E.) with respect to the horizontal in the sagittal plane.

In addition to the angle of the foot, the angle of the shank at footstrike in the sagittal plane was calculated. Results showed that at the instant of footstrike, the shank was significantly ($P = 0.03$) nearer vertical when wearing boots as opposed to shoes, and significantly nearer vertical ($P < 0.001$) when moving in a straight path as opposed to movement at a 5m radius. Shank angle was also nearer vertical in the outside leg of the curve ($P = 0.02$) relative to the inside of the curve.

	Outside foot contact time (s)	Inside foot contact time (s)	Ballistic air time (s)
Shoes straight jog	0.221 ± 0.007	0.230 ± 0.005	0.159 ± 0.006
Shoes 5m jog	0.215 ± 0.008	0.240 ± 0.008	0.087 ± 0.010
Shoes straight run	0.192 ± 0.004	0.200 ± 0.004	0.141 ± 0.005
Shoes 5m run	0.196 ± 0.006	0.217 ± 0.007	0.066 ± 0.008
Boots straight jog	0.221 ± 0.007	0.228 ± 0.007	0.155 ± 0.008
Boots 5m jog	0.218 ± 0.010	0.240 ± 0.012	0.089 ± 0.008
Boots straight run	0.191 ± 0.005	0.201 ± 0.004	0.145 ± 0.005
Boots 5m run	0.200 ± 0.006	0.213 ± 0.009	0.073 ± 0.008

Table 6.10 Table showing mean foot contact times (± S.E) and ballistic air times during straight and 5m curvilinear motion

The above table shows mean values for all conditions of straight and curvilinear motion. Most noticeable differences occurred in ballistic air time, with significantly lower time being spent between steps during curvilinear motion. For curvilinear trials, greater contact times were noted for the inside leg compared to the outside leg. Also noticeable for both the inside and outside legs were the significantly lower contact times for the running trials compared to the jogging trials.

Data from the straight trials revealed no difference between the two consecutive footstrikes, except for straight running in boots where a slight increase in contact time was seen for the second foot strike. No differences are evident when considering contact time between straight and curvilinear motion, except for the running in shoes trials where the inside foot displayed slightly longer contact time during curvilinear motion. When comparing alternative footwear conditions, no adaptations in contact time were evident. These results support the earlier work presented in chapter 4.

Trial	Shoes St Jog	Shoes 5 Jog	Shoes St Run	Shoes 5 Run	Boots St Jog	Boots 5 Jog	Boots St Run	Boots 5 Run
Gastroc medial This Study								
% off	22.00	29.28	25.14	34.90	23.34	30.29	22.36	30.85
S.E.	1.35	1.13	0.82	3.48	1.58	2.22	0.95	2.18
Gastroc medial Chapter 3								
% off	22.62	24.67	21.55	23.98	20.73	23.87	20.52	26.13
S.E.	1.03	1.12	1.22	1.03	0.78	0.69	1.26	0.85
Gastroc lateral This Study								
% off	22.48	28.70	23.91	30.47	22.89	31.60	22.14	30.91
S.E.	1.91	1.53	1.14	2.82	1.80	2.66	1.19	2.56
Gastroc lateral Chapter 3								
% off	23.53	25.01	21.83	25.81	21.16	25.37	21.19	28.42
S.E.	1.06	1.07	1.20	1.63	1.34	1.24	1.26	1.76

Table 6.11 Mean values (±S.E) for offset duration of muscular activity in the Gastrocnemius medial and lateral heads expressed as % of stride cycle. (Shaded areas show differences).

The results above show the cessation times for the activity of the gastrocnemius muscle after heel strike. The majority of conditions show activity to be similar in the present study to that reported in chapter 3 (Figures 3.2, 3.3, 3.6, 3.7). However, some conditions show different cessation of activity, with values in the present study being greater. These results may be due to alternative techniques of footstrike identification, but when differences occur they are consistent in magnitude and the movements performed in chapter 3 were considered similar to those of the present investigation.

The present study raised certain research questions which were posed as hypotheses. The results of the study have provided clear responses to those questions. It was speculated that total force values would provide some insight into the energetic transfer during curvilinear motion. Total ground reaction force values were greater at the outside leg than the inside leg of the curve, leading to the rejection of the null hypothesis and acceptance of the first experimental hypothesis. Total ground reaction force values were less at the inside leg during 5m curvilinear motion than during straight motion, which led to the rejection of both the second experimental and null hypotheses. Total force values were also considered as a summation of two consecutive strides, to negate the effect of the differing functions of the inside and outside limbs. Summed total force values were greater in straight motion than curvilinear motion which led to the rejection of both the third experimental and null hypotheses.

The mediolateral force variables considered all showed significant differences. These forces increased from straight to curvilinear motion and were greater at the outside leg. Therefore the experimental hypotheses 4 and 5 were accepted. Null hypotheses were retained for hypotheses 6 and 7 which related to foot contact time. Results from the present study verified findings of chapter 4, and confirmed that no difference in total foot contact times occurred in either the inside or outside legs, during straight or curvilinear motion.

6.4
DISCUSSION

The present investigation utilised a force platform rig for the measurement of the ground reaction force during two successive footstrikes on a natural turf surface. It was the aim of the experiment to highlight the differences between the inside and outside limb ground reaction forces, and not their modification from those of straight motion. In addition total force values were to be compared to assess differences in energetic transfer at the shoe-surface interface. The ground reaction forces were to be compared between straight and curvilinear motion at a 5m radius. Results showed significant differences in many ground reaction force variables, with values being lower for the inside leg of the curve than the outside leg.

Traces followed a pattern of classic heel-toe running described by Cavanagh and Lafortune (1980); Munro et al. (1987). A significant increase in ground reaction force was noted from jogging to running. Such results would be expected however, as higher speeds require a greater level of muscular activity to propel the body forwards (see section 3.3.1 and 3.3.2). Higher impact velocities and greater propulsive effort result in greater ground reaction forces during running.

Data from straight jogging in soccer boots gave vertical ground reaction force maximum of 2.5 *BW* ± 0.27 at 4.4 m/s. Saggini and Vecchiet (1994) reported values of 1.48 *BW* at 2.78 m/s on a natural turf surface. Saggini et al. (1992) also reported values of 1.48 *BW* but did not provide details of approach velocity or the surface used. Comparison of these data is somewhat difficult to those of the present study therefore.

Evidence from the angle of the foot at contact at the inside and outside legs, reveals differences in footstrike characteristics exist during curvilinear motion. One variable of the ground reaction force concerned with foot contact is the impact peak of the vertical ground reaction force. Significant differences existed in the time taken to reach this peak between the outside and inside limbs. However, these differences only occurred at the jogging velocity, with the outside limb taking longer to reach the peak. These differences could indicate alternative footstrike characteristics. Hamill et al. (1987) reported measures of rearfoot motion during curvilinear trials and showed levels of supination at impact were greater at the outside leg during curved motion, whereas foot contact occurred in a pronated position for the inside limb. However, the time taken to reach maximum pronation did not differ between conditions. Such results suggest that shock absorbing mechanisms may be suited to a more supinated foot position. Although the overall total rearfoot motion was greater for the inside limb (Hamill et al., 1987), the change from supination to pronation in the outside limb does relate to the shock attenuating mechanism as the mid-tarsal joint unlocks. Hence, the greater time to the first 'passive' impact peak at the outside limb. The angle of the foot at contact with the tangential line of the curve during footstrike may also affect footstrike characteristics. As the inside foot contacts with greater abduction, its mediolateral axis is more closely aligned with the direction of motion. This may cause the forward momentum of the athlete to increase the initial velocity of pronation. Any abduction of the foot at the outside limb would have an opposite effect, creating greater time to the first maximum peak as shown in the data. The differences in footstrike characteristics between the inside and outside limbs during curvilinear motion suggest that an alternative axis is used for propulsion in the two limbs. Differing axes of propulsion in the foot have previously been suggested by Bojsen-Moller (1978, cited by Viale et al., 1997).

When considering the variables concerned with the anterior-posterior (Fy) direction, statistical analysis showed the outside limb created less braking force and impulse. However, the differences in these variables were only observed at the running velocity and conflicted with those reported by Hamill et al.(1987) who showed no significant differences in any of the anterior-posterior variables measured. The absence of variation in values reported by Hamill et al. (1987) may be due to the less severe nature of the curvilinear motion monitored (radius 31.5m). Results from the present study suggest that less braking and greater propulsion occur at the outside limb during curvilinear motion. Coupled with increased vertical forces reported at the outside limb, this would suggest the outside limb is dominant in the production and maintenance of curvilinear motion. Stoner and Ben-Siri (1979) speculated the outside leg contributed centripetal force, whilst the inside leg produced an equal amount of centripetal and tangential force. Whilst the present investigation suggested the outside limb contributed a greater amount to centripetal force due to higher mediolateral forces, it would also appear to highlight a greater involvement of tangential force production (linear speed) in the outside leg. Significantly lower propulsive forces at the inside limb showed this result.

The literature review in chapter 2 suggested free moment calculations may be useful in assessing the relative contribution of the inside and outside legs to the rotatory force applied to the ground during curvilinear motion. Andrews et al. (1977, cited by Schot et al. 1995) claimed these actions were generated by torque applied to the ground, whereas Hamill et al. (1987) claimed the main mechanism was concerned with increase in mediolateral force. All values of mediolateral force showed significant differences, as the outside leg of the curve showed greatest forces. Free moment values showed that the outside leg generated greater anticlockwise moments during running. Such moments are associated with a body rotation in the direction of curvilinear motion. Results suggest that the inside leg does not contibute as greatly to the application of turning moments. Although turning moments will aid the body in rotating and maintaining a sagittal plane coincident with the curvilinear direction of travel, the mediolateral forces generated in curvilinear motion appear to be of a greater magnitude. Such findings would agree with postulations of Schot et al. (1995) that both mechanisms may be used, yet the increase in mediolateral force is the principal one.

Total force values were greatest in straight motion. Generally in curved motion the total force at the inside leg was less than at the outside leg, with the exception of the boots during jogging trials. Such results show the outside limb to provide the primary contribution to curvilinear motion. When considering total ground reaction force during straight compared to curvilinear motion, it was necessary to sum the forces over two consecutive footstrikes (Table 6.5) to gain an indication of the total force involved in a complete stride cycle. These results for the two summed total force values showed a greater overall total force to be involved with straight motion than curvilinear motion. The third experimental hypothesis suggested that the summed total force over two consecutive footstrikes would be greater during curvilinear motion. The rationale was that the same linear velocity must be maintained during curvilinear motion, in addition to generating centripetal acceleration. Results showed that greater force was required for straight motion. The reduction in overall total force could be attributed to the lower ballistic airtime during curvilinear performance (Table 6.10). Therefore a corresponding increase in vertical centre of gravity displacement during straight motion was required. These kinematic changes in conjunction with larger braking forces, accounted for the greater total force values.

Foot contact times are displayed in Table 6.10. In all conditions ballistic airtime showed a significant decrease during curvilinear motion. However, foot contact time remained unaltered from straight to curvilinear trials. The only exception occurred in running trials in shoes, where the inside limb showed increased contact time. These findings agree with those from chapter 4, where no increases in contact time were noted at a range of curvilinear grades. The decrease in ballistic airtime produces an overall shorter stride cycle, and hence a greater percentage of the stride cycle is spent in contact with the ground as the grade of curvature increases. The effect on curvilinear motion of these results would be twofold. Firstly, the greater percentage contact time allows for force application and the acceleration of the body towards the centre of the curve. Secondly, the reduced ballistic air time will ensure the athlete does not deviate further from the curvilinear path, as once the body is airborne, the centre of gravity will proceed along a path tangential to the curve.

When attempting to relate findings from earlier studies in this thesis to the present experiment, it was necessary to ensure similar movements were taking place. If patterns of muscular activity were comparable then any conclusions reached could be applied to curvilinear motion in general. Values of temporal muscular activity were compared in the two heads of the gastrocnemius muscle. Differences between the studies were evident primarily in the medial head of the gastrocnemius. The lateral head only displayed one significant difference between the experiments, whereas the medial head showed five. The lateral head results showed that the movements were comparable between the chapters. Differences occurring at the medial head could be ascribed to several factors. As differences appear consistent throughout the medial gastrocnemius data set, it may be that the technique of footstrike identification used affected results. A 10N threshold was used for identification of footstrike from the force platform, yet the binary footswitches used in chapters 3 and 4 may require greater than 10N to activate, therefore resulting in the later identification of footstrike. The earlier indication of heel strike by the force platform may account for the greater percentage activity before cessation was noted in the present study. In addition, the exact stride that contacted the platforms was used for EMG analysis in the present study, whereas chapter 3 had five seconds of EMG data from which to identify typical strides. Therefore, any irregularities in signal would be included in the analysis for the present experiment. The reason these appeared primarily in the medial head may be related to the trials in the present experiment being conducted only in an anticlockwise direction, as opposed to both directions in chapter 3. Yang (1985) concluded that EMG temporal values for some muscles can differ markedly during the gait of some subjects, however a highly repeatable kinematic pattern was associated with these changes.

6.5
CONCLUSION

A specialised force platform enabled the collection of ground reaction force data from two successive footfalls on a natural turf surface. Results from straight motion were compared to 5m radius curvilinear motion at two velocities of 'jog' and 'run' (4.4 and 5.4m/s respectively), using two footwear conditions of standard six-studded soccer footwear and flat soled indoor soccer footwear. Results showed greater total force associated with straight motion, whilst in curvilinear motion the outside leg was found to contribute most to the maintenance of the movement pattern. In curved motion, all investigated ground reaction force parameters associated with vertical ground reaction force were greater for the outside leg of the curve. Anterior-posterior forces showed the outside limb to provide greater propulsion forces and impulse. Mediolateral forces were greater during curvilinear motion. Both limbs contributed to the centripetal force associated with curvilinear motion, with forces larger at the outside limb.

The footstrikes at the platform surface were shown to deviate from the anterior-posterior axis of the plate, with foot abduction increasing with the speed of motion during curvilinear trials. For the inside leg foot abduction angles were greater, and foot contact times showed a trend to increase which highlighted differing footstrike characteristics between the two limbs. Ballistic air time was reduced from straight to curvilinear motion, creating a greater proportional foot contact time during the performance of curvilinear motion. In conjunction with lower total force values in curvilinear motion, such results suggested a lowered centre of gravity during this type of movement. A lowered centre of gravity would provide less drift towards the tangent of the curve during non-support, and a more economical transition to a position of body 'lean.' Both factors were seen as essential mechanisms for the successful progression of curvilinear motion. Foot contact time was found to decrease with increasing velocity from jogging to running yet foot contact did not alter between straight and curvilinear motion in agreement with results from earlier chapters (chapter 4).

The experiments reported up to this point in the thesis have contributed to knowledge of curvilinear motion. As indicated in chapter 1, curvilinear motion represented a reproducible and cyclical task that would enable the establishment of mechanisms underlying non-linear performance. In addition, soccer players also perform actions which contain more acute non-linear motion during the course of a game. These actions are both soccer specific and non-linear in nature, and therefore of interest here. With importance of the shoe-surface interface in soccer established by Ekstrand and Nigg (1989) and Inklaar (1994b), these soccer specific non-linear actions would provide situations where the characteristics of the interface were placed under great stress. The performance of the soccer player during non-linear motion is key to this thesis, and also to soccer footwear manufacturers. Therefore, performance of soccer specific non-linear actions would allow a potential practical dimension to the studies by examining the effect of different stud configuration on shoe-surface interaction and provides a research question for the final experimental chapter.

CHAPTER 7

CHAPTER 7

Application of a Natural Turf Force Platform Covering: Ground Reaction Force During Soccer Specific Skills and the Effect of Soccer Boot Sole Configuration

7.1 INTRODUCTION

Soccer is played on a variety of different surfaces ranging from concrete and gravel, through to natural turf and artificial grass, with each surface offering different physical properties. However, FIFA permits only the use of natural turf pitches at the highest standard of competition. Data regarding the performance of soccer players on natural turf with soccer footwear remain scarce. Only Saggini and Vecchiet (1994) have reported measurements of ground reaction forces from soccer players during straight running on natural turf. Furthermore, the ecological measurement of forces at the shoe surface interface during soccer specific movements remains absent. Some researchers (Bostingl et al., 1975) have measured the torque created by various soccer footwear on natural turf, but tests have remained mechanical in nature.

The frictional properties of a shoe-surface combination are very important to both the performance and safety of the soccer player. Inklaar (1994b) suggested the friction between the shoe and the surface can be either rotational or translational. Translational friction usually depends on the material and the structural patterns of the surface and the shoe, and is assumed independent of weight and surface area. Most tests concerning these frictional measurements involved only laboratory material testing. Van Gheluwe et al. (1983) studied frictional forces of soccer shoes on artificial turf. Three positions of 'toe-stance,' 'tip,' or 'foot stance' were analysed for 9 sole configurations ranging from flat soled indoor footwear to standard six stud soles. The authors concluded that the translational friction was greater in the foot stance position (greater surface area), hence contradicting Inklaar's (1994b) interpretation of Coulomb's law of friction. In a later study Van Gheluwe and Deporte (1992) measured different tennis sole configurations during a standard laboratory test and on court performing an open stance forehand. In the subject tests no difference was noted between the shoes, giving contradictory measurements to the laboratory tests. Such results would give support to the findings of Stuke et al. (1984), who suggested that players modify movement patterns to maintain friction within reasonable limits. Van Gheluwe and Deporte (1992) suggested that ground friction varies according to the different characteristics of the playing surface rather than those of the tennis shoes. If this notion also applies to soccer, one would postulate no difference would be observed in frictional characteristics of alternative stud configurations available in modern soccer footwear.

The moment of rotational friction depends on the pressure distribution in the contact area and the size of the contact area (Inklaar, 1994b). For injury prevention therefore, there is a need to reduce rotational friction whilst maintaining high translational friction for effective force transmission. Torg et al. (1974) attempted to classify different combinations of shoe and surface as 'safe or 'unsafe' using movements specific to American football. Those combinations of shoe and surface that gave the lowest rotational and translational friction values coincided with the lowest injury rates. However, a reduction in friction may also result in detrimental performance.

Bonstingl et al. (1975) measured the torque developed by various combinations of shoes and playing surfaces, with material tests using an artificial lower limb. Results demonstrated that a full-stance developed about 70% more torque than toe stance. When attached to the artificial limb the conventional football shoe produced the greatest amount of torque on natural grass (73.5 Nm) and also more than that generated by any other shoe-surface combination. Such results may be due to the penetrative nature of the soccer shoe causing greater fixation in the natural turf. Andreasson et al. (1986) showed mechanically developed torques of 33-43 Nm and 47-52 Nm for football shoes on artificial turf and natural grass respectively, and suggested that a balanced shoe can be achieved if a sole unit is well designed in terms of material, pattern, and frictional properties. These values are considerably lower that those of Bostingl et al. (1975). According to the definition offered by Inklaar (1994b), such design can therefore directly influence the rotational friction.

There have been few scientific research papers concerning soccer footwear. The majority of research available on soccer style footwear in the literature has emanated from American football, yet many of the findings can generally be applied to soccer. However, since American football and soccer contain significant movement differences, the shoe-surface relationship may be more critical in soccer (Ekstrand and Nigg, 1989; Inklaar, 1994b). The soccer shoe has remained virtually unchanged since the 1960's with the advent of low-cut boots and a soft leather upper. The leather upper combined with six leather, polyurethane or aluminium studs was taken as a starting point in the design of any new boot. The only variation to this design was boots with moulded sole units containing 12, 14 or 16 smaller studs for use on less compliant surfaces, giving increased surface area and pressure distribution. Monto (1993) therefore suggested that it was time for redesign, as the modern soccer shoe provides little protection, very little support and no cushioning. Cameron and Davis (1973) attempted to revolutionise the soccer boot by developing a 'swivel' shoe in an attempt to reduce ligamentous load at the joints of the lower extremity. However, the game's governing body would not allow the use of the new shoe in competition, despite the scientific evidence of its advantages by the authors. Following this, it was not until the early part of this decade that other changes became evident. A revolutionary sole unit first appeared with the Cica 'Blades' soccer boots. These comprised fin-shaped moulded studs on the forefoot and a cross-shaped heel cleat which the manufacturers, Cica, claimed would reduce stresses on the knee and be free of grass and soil build up. Unfortunately these claims were not substantiated by objective data and scientific literature was not available. Later, Adidas released a new 'Traxion' sole unit which claimed to provide greater translational friction in all directions to enhance player performance. However, once again no supporting evidence was available to substantiate such claims.

Because of the specificity of soccer movements, Ekstrand and Nigg (1989) and Inklaar (1994b) have reinforced the importance of the shoe surface relationship in soccer. The role the ground reaction force plays during curvilinear motion was classified in chapter 6, yet the effect of soccer specific movements on the ground reaction force has not been reported. It was expected that during such movements the shoe-surface interface properties would be most stressed. Therefore any changes in the shoe-surface interface properties would be elicited during such explosive, non-linear actions. The quantification of such changes represented a research question for the present study. The patterns of ground reaction force variables were to be reported during soccer specific, non-linear motions, selected to require the most extreme characteristics of the shoe-surface interface properties. The movements selected were a Cruyff turn, a drag-back turn, and a shot. The aim of the experiment was to assess the effect a modern, moulded sole unit had on the components of the ground reaction force during soccer specific movements. Such data would reflect the performance characteristics of the modern sole unit to the soccer player, and characterise the ground reaction force pattern in these non-linear actions. The experiment used the purpose designed force platform rig to measure ground reaction force during soccer specific non-linear movements on a natural turf surface.

Hypotheses

1)

H_1: There will be a difference in ground reaction force variables between moulded and studded boots during Shooting.

2)

H_2: There will be a difference in ground reaction force variables between moulded and studded boots during the Cruyff turn.

3)

H_3: There will be a difference in ground reaction force variables between moulded and studded boots during the Drag-back turn.

7.2
METHOD

Subjects

Eight male soccer players (mean age 24.4 ± 3.1 years, mass 78.3 ± 9.1 kg) volunteered for the study. Soccer players were preferred due to their familiarity with the soccer specific moves required. Experimental protocol was explained to all subjects, and all reported no musculoskeletal injuries at the time of testing. Each subject was required to provide informed consent and was reminded of their right to cease participation in the study at any time. All subjects had shoe size UK 8 or 9 and were right foot dominant.

Instrumentation

A Kistler force platform (type 9851) was mounted in a natural turf surface and placed in the first position of the cross-rig (see figure 6.1). The plate was covered with a natural turf surface as described in Chapter 6. The force plate was connected via an A-D converter to an Opus personal computer running Kistler Bioware 3.0 software. Subject velocity was measured using infra-red light gates (Cla-Win Timer, Chichester Institute, UK) placed 3m apart at hip-height on the approach to the force platform. Environmental conditions were as in chapter 6. See figure 7.1 for equipment set-up.

Video recording of the subject during each trial took place using two video recorders (Panasonic VHS Supercam AG-DP800E, EG) positioned to view the motion for reference. One camera was positioned to film foot contact with the force plate, whilst the second camera recorded the approach to assess technique and to make an audio record of the experimenter's comments.

Procedure

At the testing session, subjects were given ample time for warm up and were given sufficient practice in the testing area to adjust to the criterion approach speed and the

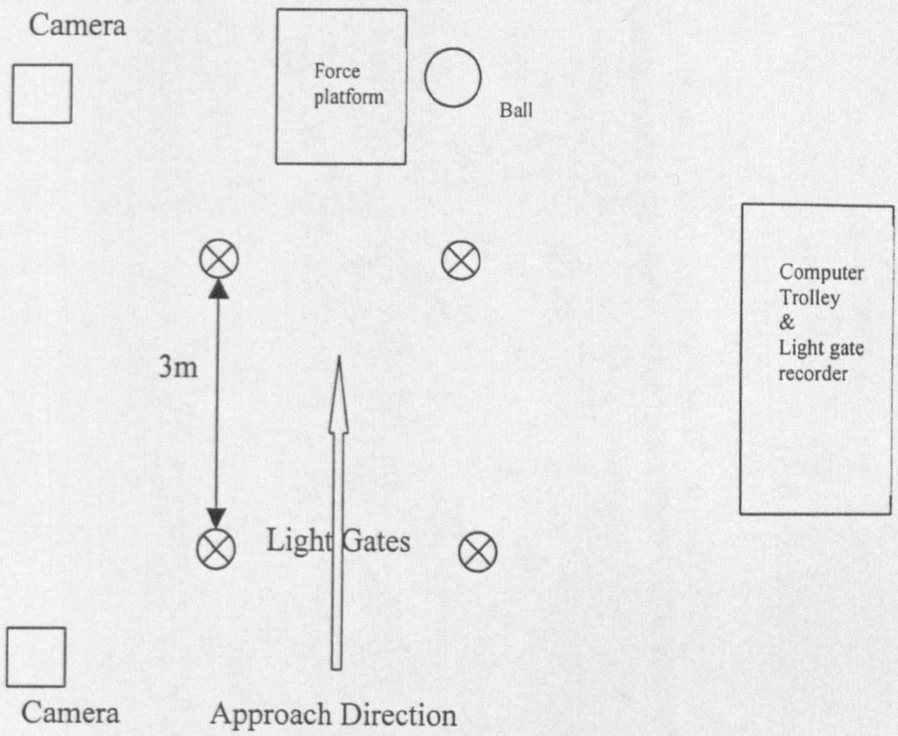

Figure 7.1 Equipment set-up.

movements required. Each subject performed three soccer specific moves, each repeated to acquire five acceptable trials. Subject approach velocity was 3.3 ± 0.3 m/s. The moves chosen were the Cruyff turn (Figure 7.2), a drag-back turn (Figure 7.3), and a shot. These were selected as those likely to create greatest frictional and torsional stress at the shoe-surface interface. Data were sampled for three seconds at 1000Hz. Turns were performed according to Football Association coaching guidelines (Hughes, 1994) with the ball placed beside the force plate, so each movement was performed with the left foot on the force platform. After completion of the turn, each subject was required to accelerate away with the ball under control, to replicate soccer match conditions. For the shot, subjects were instructed to contact the ball with the instep and strike hard and low to a target area, as if taking a penalty kick. After contact with the ball, subjects were required to land on their striking foot. Subjects repeated the series of trials in a pair of standard six-studded soccer boots (Mizuno Pro Model) and a pair of Adidas Traxion moulded sole boots (Adidas Equipment Velez Traxion), with order of footwear randomised for each subject.

Eliminated trials were those which the experimenter deemed the subject abnormally extended or shortened his stride to make contact with the platform, those in which the subject was outside of the criterion velocity, where control of the ball was lost, or where the subject deemed his performance was less than satisfactory. Generally, the subjects performed 5 to 10 trials to acquire 5 acceptable trials.

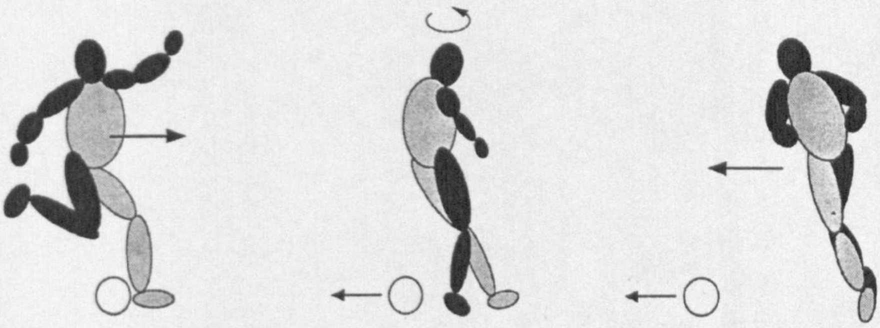

Figure 7.2 Performance of a Cruyff turn (with acknowledgement K.Sutton, UCC)

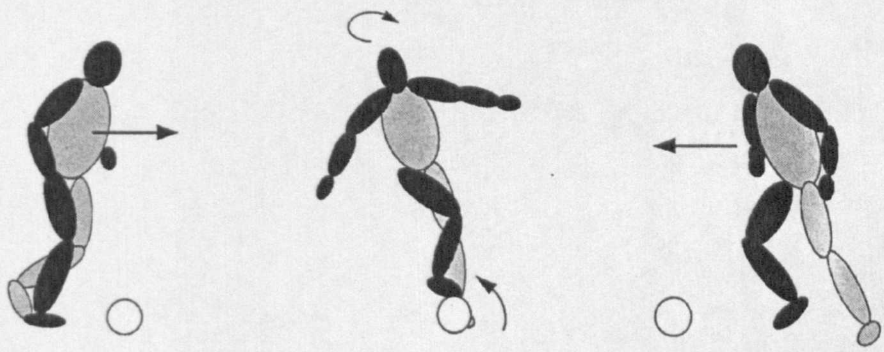

Figure 7.3 Performance of a drag-back turn (with acknowledgement K.Sutton, UCC)

Data Analysis

From the five acceptable trials, the four most technically correct were selected from the video recordings by a qualified soccer coach and analysed further. Generally the first trial was omitted from the analysis to account for the first time performance effect. Contact time with the platform was identified using a 10N threshold in the Bioware software. Vertical force, braking and propulsion forces, medial and lateral forces, free moment and torque were analysed, and normalised for each subject's body weight. Coefficient of friction in the horizontal plane was also measured, with both maximal and average friction values taken. Frictional values were computed from the modulus of the resultant force in the horizontal plane divided by the vertical force. For some trials, this computation gave large values of friction when the vertical force was low. Such large artefacts were also evident in some moment and torque patterns. Some researchers had used larger thresholds such as 10% BW, so that artefacts were minimised from division by small values of Fz in the calculation of Mz' (Holden and Cavanagh, 1991). A 10N threshold was used in the present study to gather maximum data, as the effect on Mz of studded footwear on natural grass was previously unknown.

Data were placed on a spreadsheet (Microsoft Excel 5.0) to calculate normalised force values. Mean results were calculated for each subject for each of the 12 ground reaction force variables. From these mean values, differences were computed between the two boot conditions. Standard errors were also compared to assess variance in the ground reaction force variables between boot conditions. For presentation of mean ground reaction force curves, force platform data were downloaded in ASCII file format to a spreadsheet (Microsoft Excel 5.0). A single representative subject was selected and mean traces from the four trials were generated and presented graphically. The ground reaction force variables in the vertical, anterior-posterior, and medio-lateral planes during the three movements are presented. In addition, mean values of selected ground reaction force variables were computed for each subject to enable investigation of the study objectives. Differences in mean values were then compared descriptively between the two shoe conditions in conjunction with standard error estimates. Only two conditions were used for each movement, therefore no statistical analyses were performed and raw data was presented.

7.3
RESULTS

All subjects performed the movements with sound technique. Mean graphical results for one representative subject are presented. These results were representative of each trial completed. However, some individual variation from overall mean values occurs and these are highlighted. Firstly ground reaction force variables for moulded footwear are considered, followed by studded footwear, and finally a comparison between footwear types.

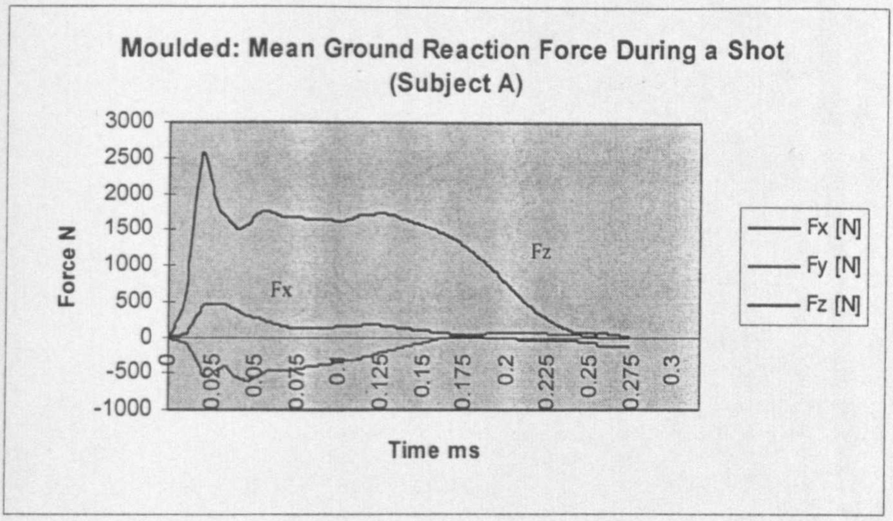

Figure 7.4 Ground reaction force during a Shot in Adidas Traxion.

N.B. Fx = Mediolateral force; Fy = anterior-posterior force; Fz = Vertical force

Vertical ground reaction force during the shot reached a maximum as the lateral border of the heel initially contacts the turf. Maximum values reached a mean of 3.74 ± 0.10 (S.E) *BW* for all subjects. The vertical trace shows the initial passive peak at footstrike, followed by a smaller peak that occurs as the striking leg is accelerated down and through the ball. An active peak is then noted as the ground reaction force serves to raise the centre of gravity through the shot and propel the centre of mass upwards in preparation for the subsequent stride. During the execution of a shot the body leaned towards the supporting leg to enable the striking foot to obtain correct position for ball strike. This position leads to a contact on the lateral border of the heel. The foot then contacted along the length of the lateral border, before pronating and directing the force towards the medial side of the boot. The Fx trace displayed this medial force throughout the stance phase as a reaction to the lateral body lean with a mean maximal force of 0.67 ± 0.06 *BW*.

The supporting leg was placed well in front of the body during preparation for the shot. The resulting acceleration in the anterior-posterior direction therefore served to slow the centre of mass. This was shown as a predominantly negative Fy trace with a maximal mean value of 1.22 ± 0.10 *BW*. The first peak coincided with initial heel contact, whereas the second peak represented the reaction to the forward swing of the kicking leg. Following contact with the ball a small positive Fy was noted as momentum carries the body forward off the platform.

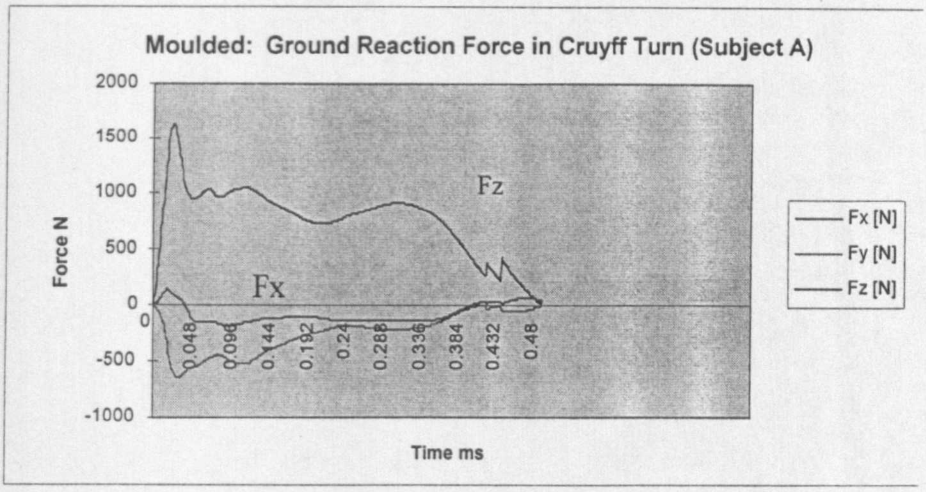

Figure 7.5 Ground reaction force during a Cruyff turn in Adidas Traxion.

During the Cruyff turn forces in the three planes showed an initial impact peak, where the lateral border of the boot makes contact with the turf. The initial peak in the Fx trace displays a positive deviation, which corresponds to initial contact with the lateral border of the foot. Pronation then occurred, which also coincided with the forward movement of the body towards the ball. This body movement however, takes the body centre of mass away from the supporting leg towards the right leg to enable the turn to be executed correctly. This movement of the centre of mass resulted in the Fx trace displaying forces directed predominantly towards the lateral border of the boot with a mean maximal value of 0.28 ± 0.04 *BW*.

The foot was placed at approximately 45° to the anterior-posterior axis of the force platform, therefore Fy values were influenced by the pronation movement as the foot first contacts the turf. The pattern of Fy trace displayed an initial peak to enable retardation of the body centre of mass, coinciding with initial foot contact. A second peak was also evident as the direction of the swinging leg was reversed towards ball contact. A third peak showed the propulsion force as the subject accelerated away from the platform. All these values remained negative with a mean maximum of 0.94 ± 0.06 *BW*, as the subject acceleration was always to reverse the direction of approach.

Vertical force (Fz) was depicted by a four peak trace. In some cases a three peak trace was noted. The maximum force was exhibited as the heel made contact with the turf and reached 2.05 ± 0.10 *BW*. The first minimum then occurred as the foot pronated to absorb some of the vertical impulse of the subject on landing. A further increase in vertical force was noted as the foot was again loaded following pronation. A third peak was evident as the swinging leg reached the limit of its forward motion and was decelerated as it passed the support leg, reversing its direction of travel to make contact with the ball. This action coincided with slightly increased knee flexion, which was observed on video records. Contact times were not normalised for the computation of mean values, therefore inconsistencies in the vertical ground reaction force trace occurred towards the end of the contact period.

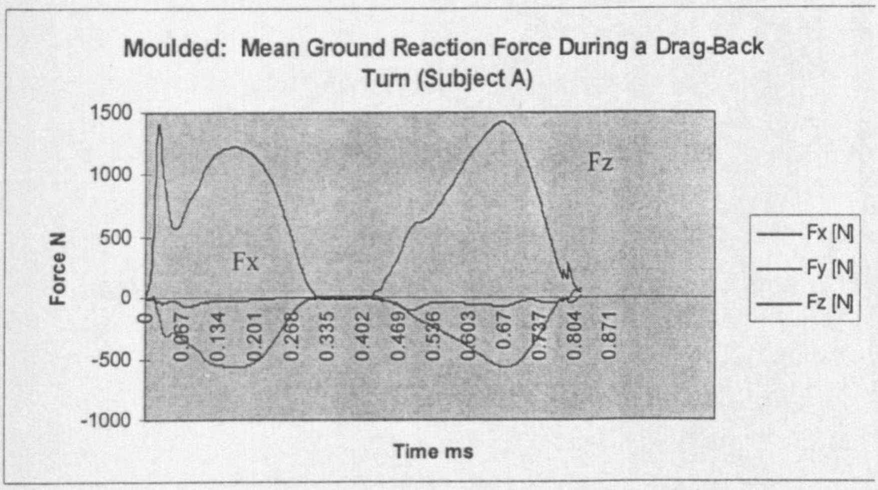

Figure 7.6 Ground reaction force during a Drag-back turn in Adidas Traxion.

The time history for the ground reaction force during the drag-back turn differs from the other movements by displaying two foot contacts. The performance of the drag-back turn requires an initial planting of the foot as the swinging leg is placed on top of the ball and then dragged backwards with the sole of the foot past the support leg. The large amount of external rotation of the support leg then requires a different foot placement for the propulsive phase of the movement.

Initial contact with the turf displayed a characteristic impact peak, as the foot contacts the turf with a standard heel-toe pattern. Mediolateral Fx values do not show the characteristic change of direction usually evident as pronation occurs. The Fx trace remains as a lateral force throughout the first foot contact, possibly due to the medial positioning of the body centre of mass with respect to the support leg. Mean maximum values were 0.13 ± 0.02 *BW.* During the second contact phase, the foot is orientated at approximately 90° to the anterior-posterior axis, with the longitudinal axis of the foot aligned along the mediolateral axis of the platform. Effectively this means that the Fy trace becomes representative of the mediolateral forces, and the Fx trace representative of the anterior-posterior forces. During this second contact, the mediolateral (Fy) forces display a large lateral force, mean maximum 0.82 ± 0.03 *BW* steadily increasing as the body is propelled away from the platform.

The anterior-posterior force (Fy) displayed a negative trace as the body centre of mass is decelerated during contact with the platform. An initial positive force was noted as the foot was swept backwards for initial contact. Such an action has also been observed during running (Cavanagh and Lafortune, 1980). The first peak coincided with heel contact, and the subsequent minimum occurred due to force attenuation from knee flexion. The active propulsion peak then served to reverse the body direction of motion and raise the centre of mass to enable replacement of the support foot with a mean maximum of 0.99 ± 0.05 *BW.* During the second foot contact, the anterior-posterior force was represented by Fx (as the foot had been repositioned at 90 degrees to the platform) with a mean maximum of 0.13 ± 0.02 *BW.* The foot was placed flat on the turf, and the direction of force transmission through the Achilles complex served to generate a negative force directed towards the rear of the foot.

The vertical ground reaction force Fz displayed a large impact peak with mean peak values of 2.06 ± 0.05 *BW.* As the curves are representative of the mean forces over four trials, one trial generated an exceptionally large initial peak, increasing the magnitude of the mean vertical force trace. The trace is typical in shape however, with the active peak coinciding with the retardation of the centre of mass. The second contact showed an initial impact attenuation spike, which is evident as a small irregularity during the loading phase. The spike has reduced in severity due to averaging of four trials, yet appears to represent the initial loading from a toe contact to a foot flat position.

The traces of ground reaction force depicted above represent the mean of those using the Adidas Traxion sole unit on a selected subject. Corresponding trials using the standard six-stud configuration in the Mizuno boot yielded similar traces. However, consistent differences in the pattern of the traces were noticeable in the mediolateral (Fx) component, possibly corresponding to the discrete medial and lateral stud placement at the heel and forefoot of the boot.

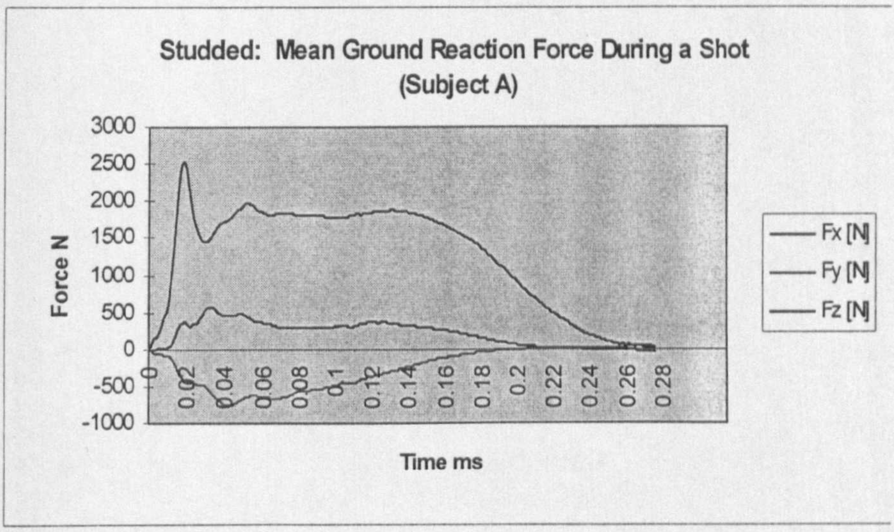

Figure 7.7 Ground reaction force during a Shot in standard six-stud sole.

	Studded sole	Moulded sole
Vertical	3.95 ± 0.14 BW	3.74 ± 0.10 BW
Anterior-posterior	1.27 ± 0.11 BW	1.22 ± 0.10 BW
Mediolateral	0.80 ± 0.06 BW	0.67 ± 0.06 BW

Table 7.1 Comparison of mean maximum forces (±S.E) occurring during the shot for 8 subjects.

When comparing the force-time histories for the shot, the most noticeable differences occurred in the mediolateral (Fx) traces. The subject depicted was representative of the general pattern. The mean maximum medial force was greater in the standard six-stud sole, as was the mean maximal braking force. In the traces presented for a single subject, the mean force-time histories show a double peak in the medial force. Such a difference was presumed to be a function of the footwear sole configuration, possibly giving an increased force loading capacity as the medial studs also interfaced with the turf.

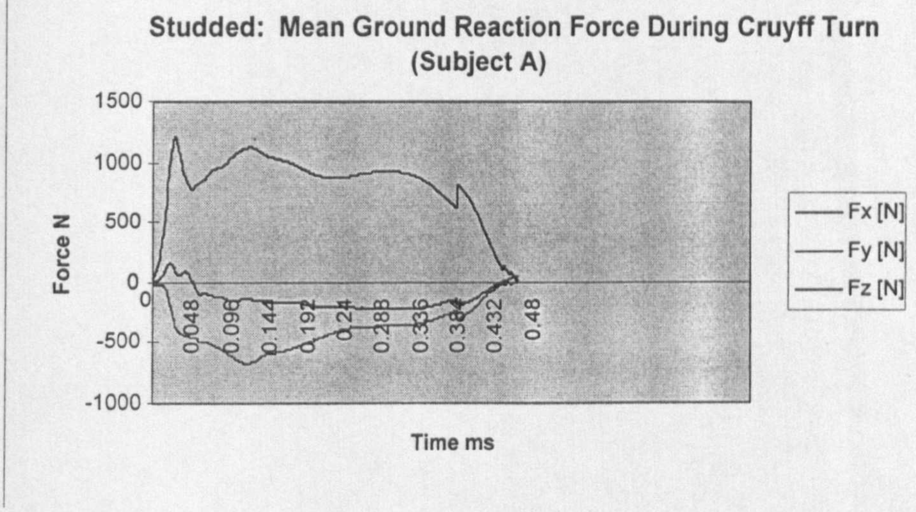

Figure 7.8 Ground reaction force during a Cruyff turn in standard six-stud sole.

	Studded sole	Moulded sole
Vertical	2.31 ± 0.15 BW	2.05 ± 0.10 BW
Anterior-posterior	1.02 ± 0.09 BW	0.94 ± 0.06 BW
Mediolateral	0.31 ± 0.04 BW	0.28 ± 0.04 BW

Table 7.2 Comparison of mean maximum forces (±S.E) in the Cruyff turn for 8 subjects.

The overall mean values for all subjects showed the maximum values of Fz to be greater in the standard six-stud sole configuration. In the Cruyff turn for subject A after the impact peak, the vertical ground reaction force appeared greater than when the moulded sole was worn. However, subject A in this instance was not typical and displayed low values for the impact phase whilst wearing the standard six-stud configuration, a trend which was opposed in the majority of subjects. Once again the double peak can be observed in the Fx trace as the lateral and medial studs contact in sequence.

Movement	Force	Variable	Ad (Mould) mean ± S.E.	Miz (stud) mean ± S.E.	Difference Mizuno-Adidas
Shot	Fy	max propulsion	0.217 ± 0.06	0.11 ± 0.01	0.100 *BW* Adidas
Shot	Fx	max lateral	-0.06 ± 0.007	-0.04 ± 0.007	0.019 *BW* Adidas
Shot	Fx	max medial	0.67 ± 0.04	0.80 ± 0.04	0.123 *BW* Mizuno
Shot	Friction	maximum	4.68 ± 0.40	3.31 ± 0.36	1.371 μ Adidas
Cruyff	Fz	maximum	2.05 ± 0.06	2.30 ± 0.09	0.265 *BW* Mizuno
Cruyff	Friction	maximum	5.27 ± 0.47	3.12 ± 0.28	2.113 μ Adidas

Table 7.3 Differences in mean values for ground reaction force variables between the moulded Adidas Traxion boot and the studded Mizuno boot for Cruyff turn and shot.

Thirty three ground reaction force variables were noted for each trial analysed. For each of these variables, differences in mean values between footwear conditions were computed. Differences in ground reaction force were evident between the standard six-stud Mizuno Pro Model and the Adidas Traxion boot and can be seen in table 7.3. During the shot the Traxion outsoles showed higher propulsive forces. In addition, the Adidas Traxion outsole showed higher maximum friction during the shot, suggesting the boot provided increased traction during shooting. The Adidas Traxion sole unit was associated with lower peak vertical ground reaction forces in the Cruyff turn, yet displayed increased values of maximal friction.

Differences between the two footwear types were noted in maximal friction values. It should be noted that these values were found at the end of the contact period, where the vertical ground reaction force was low. There were greater values of maximum friction coefficient in the Adidas Traxion outsole, but no differences between the footwear conditions when average friction coefficient over the contact period was considered.

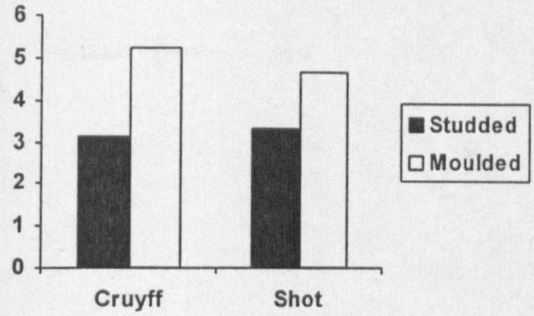

Figure 7.9 Graph showing maximal coefficients of friction in the Shot and Cruyff turn for the moulded Adidas boot and studded Mizuno boot.

	Studded sole		Moulded sole	
	Average	Maximal	Average	Maximal
Cruyff	0.57 ± 0.05	3.04 ± 0.41	0.59 ± 0.02	5.25 ± 0.47
Shot	0.37 ± 0.01	3.31 ± 0.36	0.39 ± 0.01	4.68 ± 0.40

Table 7.4 Average and maximal friction coefficient (μ) for the studded and moulded sole.

The drag back varied from the other movements in the fact that it was performed with two distinct techniques. Some subjects used one foot contact, whilst others preferred an initial foot plant to halt forward motion and initiate ball contact, and a second foot contact to accelerate away from the platform. As the two techniques enable a technically correct turn to be executed, both styles were analysed.

For subject A when comparing the two footwear conditions visually (Fig 7.6 and 7.10), it appears that a greater force is generated at the initial impact in the Traxion sole. However, when the raw data was viewed, this graphical feature was found to be caused by a large one-trial maximum affecting the subject A mean trace, and was not a typical overall mean effect for all subjects. Some of the differences observed in the drag-back turn were not easily visible from the graphical output, as their magnitude was small.

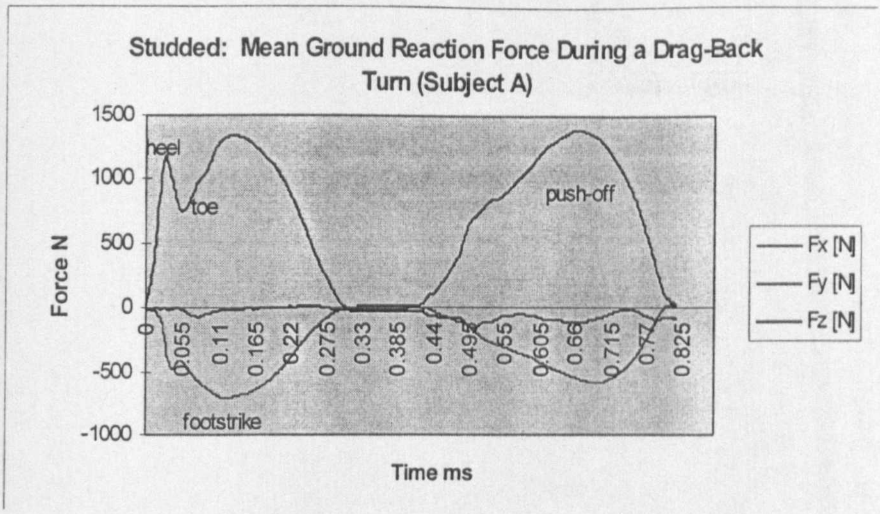

Figure 7.10 Ground reaction force during a Drag-back turn (two contacts) in standard six-stud sole.

From the thirty three ground reaction force variables noted, differences in mean values between each footwear condition was computed. If the spread of the data based on summation of the standard error values, was in excess of the difference between the means, variables were considered different. Differences were noted between the two boot types in braking and propulsion forces, and medial forces (Table 7.5). Moment, torque and friction values also displayed differences. Those who employed a two contact technique had values referred to as contact one and contact two, whereas those using only one contact had values denoted simply by the variable name.

Braking force was greater in the first contact whilst wearing the Traxion boot, which was opposite to that found in the Cruyff turn. Positive deviations in the anterior-posterior direction were few and of small magnitude, which may partly explain the conflicting results in the differences observed for this variable. Differences were also observed in the minimum (lateral) Fx value, which was greater in the Traxion boot for the second contact, and also for those subjects using a single foot contact.

Force	Variable	Ad (mould) mean ± S.E.	Miz (stud) mean ± S.E.	Difference(and greatest)
Fy	Braking contact 2	-0.82 ± 0.03	-0.67 ± 0.10	0.148 *BW* Adidas
Fy	Max propulsion	0.09 ± 0.02	0.06 ± 0.009	0.032 *BW* Adidas
Fy	Max prop contact 1	0.05 ± 0.01	0.09 ± 0.02	0.041 *BW* Mizuno
Fy	Max prop contact 2	0.10 ± 0.007	0.07 ± 0.008	0.029 *BW* Adidas
Fx	Max Lateral	-0.17 ± 0.018	-0.2 ± 0.01	0.032 *BW* Mizuno
Fx	Max Lateral contact 2	-0.13 ± 0.016	-0.31 ± 0.12	0.174 *BW* Mizuno
Mz	Maximum contact 1	0.038 ± 0.011	0.012 ± 0.003	0.026 *BW*m Adidas
Mz	Maximum contact 2	0.021 ± 0.003	0.033 ± 0.006	0.013 *BW*m Mizuno
Tz	Maximum	0.012 ± 0.001	0.017 ± 0.003	0.005 *BW*m Mizuno
Fric	Average contact 2	0.58 ± 0.08	0.46 ± 0.03	0.121 μ Adidas
Fric	Maximum contact 1	1.65 ± 0.24	2.52 ± 0.51	0.864 μ Mizuno
Fric	Maximum contact 2	4.75 ± 0.47	2.88 ± 0.29	1.877 μ Adidas

Table 7.5 Differences in ground reaction forces variables for drag back turn.

Free moment values (Mz) provide information regarding the rotational force generated through the centre of pressure during ground contact, whereas vertical torque (Tz) values give information regarding the rotational force about the vertical axis of the force platform. Inherent in turning performance in soccer is a rotational component, which may give an indication to the amount of foot fixation in the natural turf surface. Due to the nature of the movements considered, this value may also possibly relate to injury potential in the soft tissues of the lower extremity. When considering the values of free moment and torque around the vertical axis of the force platform, the free moment showed different results between the first and second contact. During the first contact the Traxion outsole gave higher moment values, whilst in the second contact, the standard outsole displayed the greatest values. There should not be too much emphasis given to such results, as the majority of the drag-back turn generated a negative (clockwise) moment. The maximum (positive) deviations were small, giving a greater likelihood of a difference emerging. For those trials where only one foot contact was used, the maximum torque about the vertical axis was greater for the standard six-stud configuration. The maximal value tended to occur with the final propulsive effort, as the heel lifted before the ball of the foot. Such a result would suggest that greater fixation occurred in the standard boot for this phase of the action. However, the general torque pattern also displayed predominantly negative values, indicated by the relatively small difference observed in positive Tz values.

The differences observed in maximal coefficient of friction occurred at the extremes of the foot contact period where the vertical ground reaction force was low. These excessive maximum friction values affected the overall mean computations and therefore were not thought to represent the majority of the movement where the foot was in contact with the turf. Although values were thought to be a valid representation of the overall contact period, the effects of these large artefacts must be considered.

The effect of these large artefacts on the mean values used was recognised. In an attempt to rectify the problems in these data, the mean values of moment, torque and friction were recalculated using a higher threshold of 100N. Such a threshold was of similar magnitude to the 10% *BW* limit used by Holden and Cavanagh (1991). This prevented the inclusion of artefacts created from calculations involving low vertical ground reaction forces. It was expected such calculations would reveal more valid differences between mean values. However, only four variables displayed differences using 100N threshold, all of which occurred during the drag-back turn (Table 7.6). Those subjects who employed a one contact technique showed higher maximal and average friction values in the standard six-stud Mizuno boot.

Variable	Ad (mould) mean ± S.E.	Miz (stud) mean ± S.E.	Difference Mizuno-Adidas
Mz Max contact 1	0.022 ± 0.005	0.014 ± 0.002	0.0086 *BW*m Adidas
Tz Max contact 2	0.009 ± 0.006	0.012 ± 0.002	0.0030 *BW*m Mizuno
Ave friction	0.45 ± 0.03	0.52 ± 0.02	0.068 μ Mizuno
Max friction	0.69 ± 0.03	0.88 ± 0.11	0.187 μ Mizuno

Table 7.6 Differences in ground reaction force variables for drag back turn using 100N threshold.

The 100N threshold did not provide a further understanding of the effect of sole configuration on the produced moment, torque and friction values. It was decided that for the analysis of the movements in the present study, the use of a 10N threshold for computation of ground reaction force variables would be most appropriate.

Overall for the soccer specific movements studied, the selected ground reaction force variables showed the standard six-studded outsole gave slightly greater values than the moulded sole. In this investigation average frictional values were similar for the moulded and studded boots during the shot and Cruyff movements. Maximal friction values showed the moulded Adidas boot to give greater values during the shot and Cruyff movements. For the drag back turn, results were mixed with the Adidas moulded boot giving greater average friction. For maximal friction, greater values were shown for the studded boot in the first contact, and the moulded boot in the second contact.

1) There were differences in ground reaction force variables between boots during the shot, therefore the null hypothesis was rejected.

2) There were differences in ground reaction force variables between boots during the Cruyff turn, therefore the null hypothesis was rejected.

3) There were differences in ground reaction force variables between boots during the Drag-back turn, therefore the null hypothesis was rejected.

7.4
DISCUSSION

This study quantified the ground reaction force of selected soccer specific movements in an ecologically sound environment. The aim was to assess the effect of a modern moulded sole configuration on ground reaction force parameters and consider results relative to those in a standard six-stud boot. Differences in these values could supply information relating to performance during these soccer specific movements. The force platform mounting rig was constructed within a natural turf surface to enable the measurement of forces whilst wearing soccer boots which penetrate the surface of the soft turf, as in match conditions. Forces were found to differ between the standard six-stud configuration and the Adidas Traxion outsole design for the three movements monitored.

Ground reaction force values obtained from the supporting leg during the shot were 3.74 and 3.95 *BW* in the vertical plane for the Adidas Traxion boot and the Mizuno studded boot respectively. These are greater than values presented by Rodano et al. (1988) in the laboratory on synthetic grass of 1.93 to 2.36 *BW*, and by Rodano and Tavana (1990) showing average vertical force readings of 2.69 *BW* for professional players on synthetic grass. Anterior-posterior forces of 1.22 for the moulded boot and 1.27 *BW* for the studded boot were also greater than the 0.88 *BW* of Rodano et al. (1988), yet were of the same order to the 1.24 *BW* reported by Rodano and Tavana (1990). Similarly, mediolateral forces of 0.6 - 0.8 *BW* measured in the present study compared with 0.5 *BW* reported by Rodano et al. (1988). The maximum vertical Fz value in the shot occurred at heel strike, therefore the magnitude of this peak should indicate the ability of the footwear to absorb shock in the heel area at impact (Bates et al., 1983). The shoe ought to be able to distribute the force so that is not concentrated in certain areas, particularly under the heel. The positioning of the studs is particularly critical in this regard, as well as the method of attachment of the stud to the boot (Lees and Kewley, 1988). A possible explanation for the differences in vertical ground reaction forces from those in the literature could be approach velocity. Although many studies did not quote approach velocities, this parameter ought to remain essentially constant for instep kicking and would not be expected to affect the ground reaction force to a large degree. The prime difference when comparing experimental conditions to those in the literature was the playing surface. The use of a natural turf covering for the force platform combined with soccer specific footwear has enabled the collection of valid data during a soccer shot, yielding values that were considerably greater than those reported previously for artificial surfaces. The forces presented for the soccer specific movements of Cruyff and drag-back turns have not been presented in the literature previously.

The shot gave differences in certain ground reaction force values between the two sole configurations. For the shot mean maximal vertical forces for all eight subjects were greater by 0.26 *BW* when the standard six studded soccer boot was worn. This was equivalent to a relative increase of 5% in vertical force when the moulded boot was worn. Most noticeable were the 16% lower horizontal mean maximal forces were in the mediolateral plane when the moulded Adidas Traxion boot was worn. These data would suggest that enough friction was generated with the moulded sole to enable greater force transmission when using this boot. There remains a possibility that subjects may not perform the shot with the same confidence in footwear markedly different from their own, and this may alter medial forces generated. Subjective confidence ratings were not monitored during the present study though comments made by subjects during the experiment did not suggest this was a factor influencing performance.

It would be expected that altered characteristics of the shoe-surface interface would affect the performance of a particular action. However, if approach speed is maintained within designated limits, and the action is performed within the limitations of the subject, one would not expect gross body movement in the performance of these selected soccer skills to differ markedly between boot types. Therefore it was not expected that large differences in the three orthogonal ground reaction force variables would be evident. However, with the two sole configurations providing different surface contact areas, one would expect altered amounts of foot fixation to be evident.

Through analysis of free moment, vertical torque values, and friction measures these features would hopefully be elicited.

The results showed greater maximal friction values for the moulded sole during the Cruyff and shot actions. During the drag-back, differences were evident for torsional and frictional properties, with generally greater friction using the moulded sole. The greater torsional values altered with each foot contact. Frictional and torsional values were both affected by large artefacts which occurred at the beginning and end of the contact period. Such a problem was overcome by Holden and Cavanagh (1991), by using a threshold of 10% *BW* to minimise the inclusion of such artefacts in the results analysis. The present study attempted to analyse the complete contact period with a threshold of 10N. With respect to obtaining smooth graphical output for moment, torque, and frictional data this level appears too low. Although the values obtained in the current study are presented, it must be remembered that they are derived from equations using the raw output from the force platform and are not direct measures of those quantities in question e.g. friction. The use of a greater threshold in the present study did reduce many of the artefacts present in the data, but such use raises questions of interpretation of reality. It could be argued that poor foot fixation generally occurs close to footstrike or close to toe-off, therefore encompassing these endpoints in any data acquisition and analysis should give a more realistic data interpretation in ecologically valid conditions. It would appear from the few differences noted when using a 100N threshold, that the adaptations, which occurred at the shoe-surface interface with different sole configurations, were at the extremes of foot contact. Therefore, for the analysis of these selected soccer movements, a low threshold for the limits of foot contact is suggested. Hence, the data discussed here was collected using a 10N threshold.

Differences in maximal friction values predominantly showed the Adidas Traxion outsole to give higher values. However, subjective responses from the subjects tended to favour the standard six-stud boot in terms of traction, sole flexibility and comfort. Therefore it would seem possible that the increased frictional values at the end of the contact phase using the Traxion outsole were possibly due to the reduced flexibility of the sole, resulting in a greater surface area of the sole being in contact as the foot left the turf. The use of the lower 10N threshold in this study enabled a further insight into these foot contact characteristics.

Compared to the values presented using mechanical tests by Andreasson et al. (1986) for studded soccer shoes on natural turf of between 0.2 and 0.9, present values were in agreement with average friction values obtained with the 10N threshold and when maximal friction data was considered at the 100N threshold. Tigermann (1983) suggested that optional ranges of translational friction for various sports, based on subjective and objective assessment, were between 0.5 and 0.7. The present study gave average friction values of 0.4 to 0.6 and much greater maximum values. However, few studies have investigated frictional properties where an outsole penetrates the playing surface. The high maximal values presented here occurred at the extremes of foot contact, where soccer players require high levels of traction to prevent slippage and loss of control. These values are therefore both necessary and specific to the particular shoe-surface interactions. In addition, the biological structures of the lower extremity are unlikely to be subjected to increased injury prevalence, as the vertical load during high frictional levels remained low. It must be realised that the mechanical tests used by previous investigations sample data slowly when compared to the 1000 Hz sampling rate used in the present investigation. Therefore, perhaps the extreme frictional values at the end of the contact period that were noted in the present study were not recorded by mechanical tests.

When comparing the rotational values obtained to those in the literature, the data reported in the present study appears to give substantially higher values for moments. During the shot, vertical free moments in excess of 60 Nm (0.078 Nm/*BW* when mean weight 768 N) occurred. This is not unexpected as torque and friction values were also higher than those obtained in other studies. Andreasson et al. (1986) showed a conventional studded shoe on natural turf generated torques between 18 Nm and 33 Nm with four different shoes using mechanical tests. Tests using subjects to quantify frictional characteristics of a surface appear to show that rotational friction is maintained below a limit of about 25 Nm by modifying movement patterns to avoid higher moments. This suggests that subjects may not predict frictional characteristics for movements that are not controlled (Nigg, 1990). However, values of 60Nm presented in the present study do appear to relate well to the 70Nm presented by Bostingl et al. (1975) for an artificial limb using strain gauges with a conventional seven-studded boot on natural turf. Bostingl et al. (1975) also reported high torque values for numerous other shoe-surface combinations, of which the lowest was 40 Nm. Mean peak values in the present study were normalised for body weight, with values reaching 0.087 Nm/*BW*, which corresponded with a mean of 67 Nm.

During the performance of the Cruyff turn, once again the average maximal friction values were greater in the Adidas Traxion boot. Also noticeable was the greater vertical ground reaction force in the standard boot. Such a result would indicate the alternative sole configuration with a larger surface area was able to dissipate the impact force with greater efficiency. Such a property would appear an advantage of the Traxion arrangement over the standard six-stud configuration.

The drag-back turn gave differences in many of the variables monitored. The Adidas Traxion outsole gave a greater braking force than the standard boot. There was also greater force directed towards the rear of the foot in the Traxion outsole (denoted by minimum Fx values). Such results would appear to indicate the superior anterior-posterior traction of the Traxion sole unit in this movement. This notion was further supported by higher overall contact 2 average friction values. A greater maximum positive free moment value was also observed with the Traxion outsole for the first foot contact. The maximum occurs at the start and end of the foot contact with the turf, with values relatively small in both shoe conditions. Values for torque about the vertical axis showed the standard six-stud arrangement to generate greater torque on the second foot contact. This coincides with the forceful propulsion away from the turn. The maximal values tend to coincide with the heel leaving the turf and the ensuing abduction of the foot through toe-off. These data would suggest that a slightly greater degree of foot fixation occurred in the six-stud design during this manoeuvre. However, values are still relatively low at approximately 5-10 Nm and do not seem to pose any direct danger to the structures of the lower limb therefore.

7.5 CONCLUSION

The force platform rig set-up enabled ecologically valid data collection during selected soccer specific movements where high forces were generated. Data was presented for forces, torque, free moment and friction during the shot, Cruyff turn and drag-back turn and showed differences in selected ground reaction force variables between a standard six-stud outsole and a modern moulded sole unit. These finding represent a novel method of data collection in sports biomechanics, and provide quantification of ground reaction force values in soccer specific non-linear motion not previously reported.

The differences in selected ground reaction force variables between the two outsole designs showed that the Traxion outsole gave consistently higher friction values. When considering the three orthogonal ground reaction components, this plastic bladed sole unit also showed reduced vertical ground reaction forces during the maximum impact phase of the Cruyff turn, and overall lower forces during the shot.

It has been suggested that subjects may alter their movement pattern in response to use of different footwear (Stuke et al., 1984) a problem inherent in subject based experiments. The present study focussed on modifications of conditions at the shoe-surface interface, with measures taken at the interface in the form of ground reaction force variables. Whilst alteration of gross body movement may have occurred in response to altered sole characteristics, quantification of such movement was beyond the scope of the present investigation. However, future inquiry into the effect of altered sole configurations should consider the study of the relationships of footwear to whole body movement, and also the assessment of ground reaction force variables during realistic cutting movements which generate amongst the highest forces encountered during soccer performance. The controlled movements investigated here elicited some changes in ground reaction force variables, yet unexpected movements for which the body cannot prepare may provide the main dangers to injury in the soft tissues when characteristics at the shoe-surface interface are modified.

Overall, results show that a natural turf covered force platform can be used to collect data for soccer specific non-linear motion. These data can provide an insight of the mechanisms involved in such movements. Results support the findings of chapter 6 where mechanisms of both an increase in ground reaction force and increase in turning moments were responsible for the generation and maintenance of non-linear motion. Data interpretation can also highlight the effect of differing sole characteristics on the forces experienced at the shoe-surface interface. For the present investigation computed vertical torque and free moment values showed differences in values during the drag back turn, yet values were low in magnitude as they were opposite in direction to the rotation occurring during the turn. These results suggested that the moulded sole with its larger bladed contact area allowed these three soccer specific movements to be performed while exposing the body to reduced levels of ground impact. This property of the moulded sole, with its blade configuration should reduce the risk and severity of impact related injury and appears advantageous over the standard studded sole. The results indicate that the average coefficient of friction was similar for both soles while the moulded sole boot has the ability to allow greater maximal friction at the beginning and end of the ground contact when slippage is most likely.

CHAPTER 8

CHAPTER 8

DISCUSSION AND CONCLUSION

8.1
GENERAL DISCUSSION

The aim of this thesis was to establish the mechanisms of non-linear motion specific to soccer performance. The area was devoid of previous scientific enquiry, although a few investigations had occurred into applicable actions such as cutting movements (Schot et al., 1995) and curvilinear motion in athletics (Greene and McMahon, 1979; Stoner and Ben-Siri, 1979; Greene, 1985; Hamill et al., 1987). This research aimed to investigate the mechanisms that enable humans to move in a non-linear path, and to relate these to performance in soccer. As soccer is generally played on a natural turf surface, methodological problems had to be overcome to enable the collection of data in an ecologically valid environment. It was thought that the analysis of curvilinear motion would provide information on the basic mechanisms causing non-linear motion, within an easily reproducible and standardised activity.

The alternative path of movement during non-linear motion was thought to emanate from differing muscular activity in the lower extremity. Initial investigation into curvilinear motion therefore involved quantification of this muscular activity, with the aim of highlighting the subtle differences that enable curvilinear movement to occur. From EMG studies into barefoot and shod running (Stockton and Dyson, 1998), differences between footwear conditions were reported. Thus, EMG was used to investigate differences at the shoe surface interface during curvilinear motion. Subsequently data were collected at differing grades (radii) of curve, athlete velocity, and shoe-surface interface. The muscles monitored were those which were found to control movements of the thigh, shank and foot in the frontal plane. Results showed that the magnitude of muscular contraction did not differ between grades of curve, but that values increased with greater speed of locomotion. Curvilinear motion performance was associated with temporal adaptation of muscle activity at all grades of curve but especially at the 5m radius, due to the greater proportional increase in curve severity. The radii of the three curves were 15m, 10m and 5m respectively. However, the 5m radius has twice the severity of the 10m radius, yet the 10m radius has only 1.5 times the severity of the 15m. The radius has a direct effect on the centripetal force required for curvilinear motion (Chapter 1, p.15). Muscular adaptation was evident in both legs, although predominantly in the outside leg of the curve, and took the general form of prolonged activity after heelstrike. Increased muscle activity around stance would enable the prime muscles for curvilinear progression increased time to position body segments. These preliminary investigations also showed that stride kinematics were altered during curvilinear motion, with an increase in stride frequency and a decrease in stride length as the curve became more severe. The research described in chapter 4 showed no increase in foot contact time at any of the grades of curved motion. However, due to the reduced stride length and increased stride frequency, an increased proportional foot contact time as the grade of curve became more severe was evident. Such an adaptation gave greater time for the musculature of the lower extremity to apply force to the ground during curvilinear motion, and is a key finding of the analysis. Such results therefore explain the increased muscle activation time after heel strike reported in chapter 3. Curvilinear motion requires a mediolateral movement, which was controlled by the muscular activity around the ankle.

It was expected that the effect of soccer footwear upon curvilinear performance would be reflected in the electromyographical data, and the data would enable the distinction between different frictional properties of an alternative shoe-surface interface. Data from chapter 3 showed alterations in muscle amplitudes only coincided with temporal differences when the subjects were wearing soccer boots (Tables 3.8, 3.9 & 3.10). The changes occurred in the tensor fascia latae during swing, and gluteus maximus muscles during stance, indicating that these muscles were important in enabling muscles in the outside leg to achieve curvilinear motion at a 5m radius. However, these could not be considered the prime movers for such actions as no recordings were made from muscles such as iliopsoas or the hamstring group in the present investigations. Peroneus brevis showed a significantly later offset of activity after heel strike in the inside leg, which indicated differing footstrike characteristics between the inside and outside limb during curvilinear motion. The electromyography study made a significant contribution to the understanding of adaptation of the muscles of the lower extremity during curvilinear motion (Smith et al., 1997). Also, research has shown that lower extremity EMG measures can be used to identify differences at the shoe-surface interface.

Temporal muscle adaptations were thought to apply force to the lower extremity in a way to cause altered segmental accelerations during curvilinear motion. Such adaptations were expected to reveal altered stride kinematics. Chapter 5 reported the investigation into the kinematics of the lower limb during curvilinear motion, revealing adaptations in angular displacement measures with increased severity of curve. Adaptation occurred primarily at the inside leg of the curve as flexion at the joints was increased during curvilinear performance. Such results can be explained by reference to the modified stride pattern. A shorter stride length would provide less vertical oscillation of the centre of gravity, and therefore greater joint flexion as the body position is not so upright. A lower centre of gravity would be mechanically beneficial by reducing the turning moment of the body in the frontal plane. This lower position would also require less lateral movement of the centre of gravity to attain the body 'lean' required for maintenance of curvilinear motion. Asymmetry of movement during curvilinear motion was characterised by greater knee flexion at the outside leg during support. In addition, ankle movement was different in each leg, which supported the findings of earlier work in muscle activity (chapter 3) by highlighting the ankle to be the site of key adaptation. The different footstrike characteristics arising from chapter 3 were highlighted by no change in inside leg ankle angle with increased curve severity, but a greater angle at the outside leg ankle. The findings of the lower centre of gravity and the ankle as a key site for adaptation were new and add to the understanding of non-linear motion. Although the sagittal plane adaptation was noted at both ankles, it is probable that a holistic mechanism for curvilinear motion would involve rotation of the limbs also. Unfortunately, experimental and software limitations did not allow the capture of such data, although the values presented do provide an indication of the adaptations that are occurring during curvilinear performance.

The asymmetric kinematics reported in chapter 5 indicated a different role for the inside and outside leg of the curve. Data from chapter 3 showed greater muscular adaptation at the outside limb. These combined findings were strengthened by ground reaction force results in chapter 6. During curved motion, all investigated vertical ground reaction forces were greater at the outside leg. The inside limb showed less propulsive force and impulse. Both limbs contributed to the maintenance of centripetal force required for curvilinear motion by increased mediolateral forces, with the outside limb again displaying greater values. Therefore, the outside limb made a greater contribution to the mechanism of curvilinear motion, which was a key finding of the research. Greater forces would be expected at the outside leg as curvilinear motion provides a tendency for the body to continue on a tangential path, and therefore to move toward the outside of the curve. As the outside leg is positioned furthest from the body centre of gravity, it must create a greater centripetal force to maintain curvilinear motion. Force is then absorbed due to the centre of gravity movement toward the outside of the curve. The increased mediolateral force component is then generated as a function of the body 'lean.' The greater force absorption (also indicated from larger knee flexion) enables storage of elastic energy in the structures of the outside leg, providing greater force and impulse during propulsion. Contrasting results were provided by Hamill et al. (1987), who showed no differences in anterior-posterior forces during motion at a 31.5m radius. Their non-significant findings may be due to the larger radius used, yet the relative contributions of the inside and outside limb to propulsive forces requires further investigation.

Once again differing footstrike characteristics were evident between the two limbs as the inside limb showed greater abduction of the foot. The collection of force platform data enabled very accurate measurement of foot contact time in straight and curvilinear motion. Results showed that foot contact time did not change during curvilinear motion, supporting findings reported in chapter 4. These findings show a reduced ballistic air time during curvilinear motion, which gave a greater percentage of the stride cycle in contact with the ground. Also total force values, summed over two successive footfalls were shown to be greater during straight than curvilinear motion. These combined results would serve to reinforce the view that an important mechanism in curvilinear performance was the maintenance of a lower centre of gravity (p216). If this can be achieved, then the athlete can reduce ballistic air time between footstrikes by directing a greater proportion of the total force along the mediolateral axis. This will provide the centripetal acceleration required for curvilinear progression. When the body is in periods of non-support, physical principles tell us that the centre of gravity will move in a direction tangential to the curved path of motion. Therefore, the reduced ballistic air time also serves to minimise any deviation from the curvilinear path that would occur during this time. This rationale provides a key mechanical explanation for motion in a curvilinear path, which contributes to knowledge in the area of non-linear motion.

A combination of findings showed that increased shock attenuation mechanisms are evident at the outside leg of the curve. The effect of different footstrike characteristics was evident in the ground reaction force data measures. Increased time to the first maximum impact force was noted in the outside leg whilst jogging. Such a result can be related to a decreased outside leg ankle angle at greater curve severity, which would indicate a greater amount of dorsi flexion at impact. In addition, Hamill et al. (1987) reported that the foot of the outside leg during curvilinear motion showed a greater amount of supination at heelstrike. Also, an increased amount of knee flexion during stance was reported at the outside leg in chapter 5. These combined mechanisms possibly occur in response to the greater forces generated by this limb. Findings suggest that eccentric strength training for the gluteal and quadriceps muscle groups may aid in the attenuation of force to the outside leg. Also, such training may reduce the amount of knee flexion noted during support enabling a faster and more powerful motion.

The main characteristics of curvilinear motion have been highlighted in chapters 3, 4, 5 and 6, yet many other types of non-linear motion are performed during a game of soccer. Sports specific movements such as the shot and turning create large forces at the shoe-surface interface, with the effective transmission of these forces to the ground essential for good performance. In recent years several new sole designs have been manufactured in an attempt to improve performance. The force platform rig designed for measurement of ground reaction forces on natural turf was used to quantify the forces created during such movements. In addition, measurements of the rotation forces and friction provided a greater insight into the effect altered shoe-surface characteristics has on performance. Chapter 7 reported ground reaction force variables during a shot, drag back turn and Cruyff turn using a standard six-stud outsole and a modern moulded sole unit. Stuke et al. (1984), suggested that players modify movement patterns to maintain friction within reasonable limits. If such speculation were true, the use of different outsoles would not provide differing ground reaction force variables. However, results showed that the moulded Adidas Traxion outsole gave greater frictional values in the performance of the actions chosen. In addition, this boot showed lower vertical forces during the maximum impact phase of the Cruyff turn, and overall lower forces in the shot. Such results have implications on performance of these movements and provide a practical application to the biomechanical analysis of non-linear motion in soccer. The study aimed to measure the effects of altered shoe-surface conditions on ground reaction force variables during soccer specific non-linear motion. The suggestion that force values would not vary in response to differing sole conditions was refuted, with decreased impact forces occurring in the modern, moulded sole. Such a finding has contributed to knowledge by providing a method for comparison of soccer sole units in a sound ergonomic environment. Tests used human subjects, on a natural turf surface, performing soccer specific actions. In addition, frictional characteristics can be elicited through measures of ground reaction force on a natural turf surface, and showed increased values for the modern, moulded sole.

Several limitations of the studies reported have become evident. These are predominantly concerned with the practicality of subject based experimentation. Increased subject numbers would obviously provide greater statistical power to the results, yet must be balanced with the increase in experimentation and data analysis time. The previous research published in the area was difficult to find. Following the research into EMG, cross citation revealed earlier studies relating to non-linear motion (Stoner and Ben-Sira, 1979; Hamill et al., 1987; Schot et al. 1995). However, results from the initial EMG investigation raised further questions, and the progression of the thesis research had been established. Soccer also contains a vast array of different skills, many of which rely on patterns of non-linear motion. Unfortunately, with the paucity of literature in the area at the outset of the thesis, the basic mechanisms of curvilinear motion needed to be ascertained before soccer specific movements could be addressed. Once basic mechanisms into curvilinear motion had been investigated, applications to soccer specific movements were considered.

However, the limitations of the studies reported do provide impetus for future research in the area. The quantification of the three dimensional nature of motion in a curved path remains unreported in the literature. Specifically, research needs to address the differences in internal and external rotation of the lower extremities and torso. Such work would also aid in description and quantification of differing footstrike characteristics between the inside and outside limbs during curvilinear motion. Although the role of body 'lean' was understood with respect to counteracting rotations in the frontal plane, the quantification of the 'lean' during curvilinear motion would be of interest. Knowledge of the position of the centre of gravity during curvilinear and soccer specific actions would strengthen the suggested mechanisms involved in these movements.

The vast array of soccer specific moves also provides increased opportunity to investigate the aspects of technique which enable successful performance. As was seen from results in chapter 7, the application of biomechanics in the evaluation of shoe-design can provide important results. Future research might to be directed towards the optimisation of the soccer boot sole. New designs can be assessed using human subjects on a natural turf surface. As changes in interface characteristics seemed to modify kinematics (Stuke et al., 1984; Nigg, 1990), the combination of force platform and simultaneous kinematic measures would obtain data useful in attempts to increase effective force transmission, whilst maintaining low risk to the soccer player.

8.2
CONCLUSIONS

Investigation into the biomechanics of non-linear motion has revealed several novel findings, which aid our understanding of curvilinear and soccer specific movements. Initially, research centred on muscle activity during different grades of curvilinear motion and established temporal differences in activity pattern, and also detected changes in muscular activity when training shoes were compared to soccer boots. Temporal muscle activity displayed adaptation as the grade of curvature became more severe. Greater adaptation in the outside leg of the curve was evident, with muscle activity tending to increase in duration around the stance phase. Changes in muscle activity pattern between different types of footwear showed selected muscles to have delayed activity patterns when boots were worn. Again these patterns were most prevalent in the outside leg of the curve. Alterations in stride kinematics were also noted in curvilinear motion as stride length was reduced, and stride frequency increased with curve severity. Chapter 4 verified these kinematic measures and showed that an increase in the proportion of the foot contact time was evident. These results could, in part, explain some of the EMG adaptations noted in the earlier work (chapter 3). The musculature of the lower extremity had a greater time to apply force to the ground during curvilinear motion, and therefore alter the acceleration of the segments of the lower extremities. Results showed that curvilinear motion contains a mediolateral component, which is controlled by muscular activity around the ankle joint. Muscular activity during non-linear motion had never been reported prior to these investigations, and therefore provides a unique contribution to knowledge.

Adaptation to curvilinear motion was noted with the division of foot contact time between forefoot and rearfoot. A tendency for increased rearfoot time as the grade of curve became more severe was seen in the outside leg of the curve in soccer boots. In conjunction with differing muscular activity around the ankle in the inside and outside leg, these data indicated altered footstrike characteristics between the two limbs in curvilinear motion.

Originally, the alternative path of movement in curvilinear motion was thought to emanate from altered muscular activity. When the kinematics were investigated, greater flexion occurred during curvilinear motion, specifically at the inside leg. Shorter stride length and increased joint flexion provided a lower body centre of gravity, which reduced the tendency of the player to rotate over the feet in the frontal plane, towards the outside of the curve. To combat the turning moment, players would 'lean' toward the centre of the curve. The lowered centre of gravity also gave less distance to move to create the required 'lean.' The proposal of the mechanism for altered stride kinematics to aid curvilinear motion represented a contribution to the understanding of non-linear motion. Kinematic analysis again highlighted the ankle as a key site for curvilinear motion, with a difference in function of the foot/ankle complex indicated. Lower extremity kinematic responses to straight running were found to agree with those published in the literature. Overall, ranges of motion were shown to decrease during curvilinear motion as stride length was reduced.

Ground reaction forces on a natural turf surface with standard soccer footwear were investigated to quantify the different contributions of the two limbs. Curvilinear motion was associated with greater vertical ground reaction forces at the outside leg. Both legs were shown to contribute to the centripetal force by increased mediolateral forces, yet the greater magnitude was again noted for the outside leg. A key finding arose therefore, as the outside leg was shown to provide a greater contribution to curvilinear progression.

Foot contact times obtained during the ground reaction force study verified earlier work (chapter 4) and also showed ballistic air time to decrease during curvilinear motion. Decreased ballistic air time also served to lower the vertical oscillation of the body centre of gravity. Less time spent in non-support between strides also reduced the drift of the body along a path tangential to the curve, and therefore aided a smoother curvilinear path. Chapter 6 provided insights into the total kinetic transfer at the shoe-surface interface during curvilinear motion. Summed force over two consecutive strides showed curvilinear motion at 5m radius gave lower total force values than straight motion. Explanation was referenced to the requirement for less vertical force to raise the centre of gravity, as shorted strides were taken during curvilinear motion. Results suggest that the muscular adaptation, coupled with altered stride kinematics provide an economical mode of progression through a curvilinear stride cycle.

Differences in sole configuration were investigated at the shoe-surface interface. Measures gave an application of ecologically valid force platform measurements in soccer specific non-linear motion. Results noted that a modern moulded Adidas Traxion outsole gave greater friction and lower vertical ground reaction forces when three movements of shot, Cruyff turn, and drag back turn were performed. The results have implications on the performance of these movements. With greater frictional values toward the extremes of foot contact, the modern sole can reduce slippage during initial foot contact and final propulsion, enabling increased performance. Lower vertical ground reaction forces would decrease loading on the musculo-skeletal system, and may therefore reduce impact related injury prevalence. In addition, an important finding was that ground reaction force reflected differences between footwear types under conditions that were relevant to the game of soccer. Such a method of footwear assessment can provide useful data to manufacturers and clinicians concerning the potential of sole unit designs for effective force transmission during sports specific situations. Such an experimental set-up is suggested by the author of this thesis for future ergonomic evaluation of soccer outsole designs.

In conclusion, the research detailed within this thesis provided new evidence concerning the mechanisms involved in the performance of non-linear motion in soccer. The initial study demonstrated some of the altered muscular adaptation in the lower limb which was the mechanism by which the body adapted its function to follow a curved path. The muscular adaptations primarily occurred in the form of increased duration of muscular activity around the stance phase. These muscular adaptations resulted in changes to stride kinematics, with greater flexion evident at the inside leg of the curve. Stride length decreases, stride frequency increases, proportional stance time increases and increased leg flexion at the inside leg have been identified in response to the need to perform curvilinear motion. Different adaptation responses have also been identified with greater forces transmitted at the outside leg of the curve.

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APPENDIX A

(Relevant to Chapter 3)

For **each muscle** activity is listed during jogging and running, initially for the boot condition and training shoe condition in curved motion
Following this, activity in straight motion in boots and training shoes

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

Tibialis Anterior

key:

off = cessation of activity
dur = duration of activity
on = restart of activity
SE = standard error

	Boots											
	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	19.73	19.56	17.93	20.76	19.73	17.59	22.29	20.44	19.77	22.44	18.83	20.58
ave dur	19.73	19.56	17.93	20.76	19.73	17.59	22.29	20.44	19.77	22.44	18.83	20.58
ave on 2	27.34	29.55	26.85	26.03	28.63	27.51	27.93	28.43	27.58	29.27	26.43	27.10
ave off 2	72.08	72.95	68.70	69.88	69.70	72.22	71.17	69.72	70.14	71.64	67.84	71.44
ave dur 2	44.74	43.38	37.39	40.46	41.54	44.73	41.49	41.29	42.57	41.91	41.41	44.33
ave on 3	79.84	78.76	78.34	77.51	78.43	77.06	82.19	80.21	79.01	80.42	78.93	79.15
SE off	1.30	1.01	1.21	1.64	1.40	0.94	1.02	0.82	0.89	0.96	0.89	0.88
SE Dur	1.30	1.01	1.21	1.64	1.40	0.94	1.02	0.82	0.89	0.96	0.89	0.88
SE on2	1.89	1.86	1.69	1.93	2.92	2.15	1.42	1.69	2.08	1.80	1.10	1.49
SE off2	2.73	1.32	1.88	1.90	1.65	1.62	2.41	2.18	2.11	1.94	2.09	2.25
SE Dur2	2.87	2.62	2.67	3.69	3.20	2.62	3.61	2.66	2.52	1.68	2.20	2.86
SE on3	1.44	1.26	1.87	2.50	1.29	1.60	2.60	2.34	1.61	2.26	2.35	2.20

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

Tibialis Anterior

	Training shoes											
	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	20.31	21.29	18.79	19.72	20.71	17.67	21.76	19.86	19.39	21.42	20.99	19.40
ave dur	20.31	21.29	18.79	19.72	20.71	17.67	21.76	19.86	19.39	21.42	20.99	19.40
ave on 2	27.91	29.24	27.55	28.50	31.02	25.61	29.12	28.40	28.43	28.13	27.29	26.86
ave off 2	70.93	72.41	68.59	72.20	72.82	68.46	67.24	72.33	69.73	69.58	66.60	66.03
ave dur 2	42.29	43.00	40.91	42.29	45.03	42.89	40.39	43.92	41.27	39.65	42.03	37.66
ave on 3	76.37	75.73	74.12	79.05	76.79	77.40	77.08	78.68	77.11	80.40	75.83	74.36
SE off	0.82	0.89	1.05	1.24	0.88	0.74	0.89	0.65	0.69	1.19	1.07	1.11
SE Dur	0.82	0.89	1.05	1.24	0.88	0.74	0.89	0.65	0.69	1.19	1.07	1.11
SE on2	1.88	2.20	1.78	2.06	2.96	1.98	1.13	1.58	1.67	1.67	1.71	1.55
SE off2	2.20	2.23	1.76	1.07	1.86	1.85	4.63	1.85	2.47	1.82	3.69	2.76
SE Dur2	2.93	3.24	2.55	3.53	1.99	2.69	4.23	2.72	2.20	2.88	2.27	2.84
SE on3	1.95	2.09	2.10	1.83	2.10	2.07	4.61	2.49	2.80	1.54	2.82	3.26

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

Tibialis Anterior

	Boots		Training shoes	
	jog straight	run straight	jog straight	run straight
ave off	19.73	18.62	19.52	17.18
ave dur	19.73	18.62	19.52	17.18
ave on 2	29.89	26.60	28.29	24.98
ave off 2	73.60	68.57	71.21	66.16
ave dur 2	43.32	41.97	42.91	41.01
ave on 3	78.13	76.03	78.76	75.19
SE off	1.18	0.87	0.82	0.68
SE Dur	1.18	0.87	0.82	0.68
SE on2	2.15	2.02	1.66	1.35
SE off2	1.71	1.45	1.46	1.31
SE Dur2	2.77	1.85	2.31	1.18
SE on3	1.93	2.11	1.46	1.71

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

Peroneus Brevis

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

	Boots											
	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
Ave off 1	25.30	24.33	22.77	26.12	23.89	23.06	23.14	22.92	22.06	20.67	22.37	21.03
Ave Dur 1	25.30	24.33	22.77	26.12	23.89	23.06	23.14	22.92	22.06	20.67	22.37	21.03
Ave on 2	34.13	40.63	29.95	35.05	33.40	29.50	27.50	28.07	26.77	30.46	29.04	27.50
Ave off 2	58.57	67.67	64.30	64.00	53.90	61.35	52.68	55.23	57.43	70.26	55.07	53.53
Ave Dur 2	24.37	26.97	34.35	28.95	20.47	31.88	21.23	27.17	30.67	39.78	26.06	26.04
Ave on3	68.50	79.40	69.60	74.85	62.10	68.65	65.10	49.20	47.60		60.70	56.90
Ave off 3	89.25	90.50	83.00	94.80	87.90	81.55	79.20	77.80	81.00		79.15	81.00
Ave Dur 3	20.75	11.10	13.40	16.27	25.90	12.90	15.50	28.60	33.30		18.45	24.10
Ave on 4	85.83	86.55	84.27	78.15	81.43	86.75	83.37	84.36	85.10	81.53	82.13	82.39
SE off 1	2.09	1.61	1.80	1.70	1.50	1.55	1.02	1.17	1.38	2.36	1.61	1.28
SE Dur 1	2.09	1.61	1.80	1.70	1.50	1.55	1.02	1.17	1.38	2.36	1.61	1.28
SE on 2	1.07	4.02	2.39	1.83	2.43	1.24	0.82	0.36	0.53	2.88	2.62	0.85
SE off 2	0.67	3.56	0.18	1.35	2.33	4.00	3.46	2.86	3.65	2.59	2.99	2.46
SE Dur 2	0.53	1.29	2.21	3.17	4.71	4.03	2.94	2.56	3.13	4.48	2.65	2.73
SE on 3	1.93		4.16	3.42		4.94	3.58				1.70	
SE off 3	0.26		4.47	0.65		6.15	0.63				0.07	
SE Dur 3	2.19		0.31	2.22		1.21	2.28				1.77	
SE on 4	2.41	2.35	2.09	3.77	2.61	2.14	2.36	2.11	1.07	1.24	2.46	1.27

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

Peroneus Brevis

	Training shoes											
	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
Ave off 1	25.67	25.88	23.14	27.00	25.43	23.05	23.19	23.37	21.77	22.99	26.23	24.44
Ave Dur 1	25.67	25.88	23.14	27.00	25.43	23.05	23.19	23.37	21.77	22.99	26.23	24.44
Ave on 2	37.88	35.55	34.83	39.40	32.08	31.40	31.88	32.15	30.57	30.27	34.84	34.82
Ave off 2	69.08	68.03	63.30	61.45	64.48	61.80	61.30	61.50	50.97	53.68	66.62	65.22
Ave Dur 2	31.20	32.48	28.47	22.05	32.58	30.48	29.40	28.35	20.37	23.40	28.56	30.10
Ave on3	75.00		74.75	77.17	75.40	76.10	57.40		50.00	54.50		
Ave off 3	93.30		94.50	94.73	97.40	91.00	75.40		83.90	74.50		
Ave Dur 3	16.70		19.20	17.63	22.00	14.90	18.00		33.90	20.00		
Ave on 4	82.74	81.30	85.89	91.03	87.54	81.70	81.86	85.39	85.84	78.61	84.22	84.14
SE off 1	1.66	1.69	0.65	0.62	0.95	1.30	0.50	1.40	1.75	1.35	2.07	2.46
SE Dur 1	1.66	1.69	0.65	0.62	0.95	1.30	0.50	1.40	1.75	1.35	2.07	2.46
SE on 2	3.04	1.58	2.97	5.67	1.60	1.49	2.43	0.96	1.35	1.66	0.53	1.90
SE off 2	2.00	2.10	1.81	4.09	2.06	1.12	5.40	0.80	4.03	4.91	1.66	2.59
SE Dur 2	3.39	0.91	1.29	1.58	3.36	2.26	3.43	2.21	3.12	3.63	1.65	3.12
SE on 3			0.25	1.27	0.98							
SE off 3				1.12	0.36							
SE Dur 3				0.61	1.34							
SE on 4	1.75	1.65	1.93	4.86	3.09	2.42	1.65	0.92	1.99	2.47	1.43	1.66

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

key:

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SE = standard error

Peroneus Brevis

	Boots		Training shoes	
	jog straight	run straight	jog straight	run straight
Ave off 1	20.14	23.48	22.26	19.90
Ave Dur 1	20.14	23.48	22.26	19.90
Ave on 2	34.98	29.54	31.03	32.48
Ave off 2	61.50	60.26	60.03	60.70
Ave Dur 2	26.50	30.75	29.00	28.25
Ave on3	66.78	79.35	49.20	
Ave off 3	86.58	97.55	87.30	
Ave Dur 3	19.80	18.15	38.10	
Ave on 4	85.70	85.98	87.33	81.29
SE off 1	0.99	1.77	1.07	1.95
SE Dur 1	0.99	1.77	1.07	1.95
SE on 2	2.40	1.74	2.34	1.85
SE off 2	0.77	3.85	1.76	2.33
SE Dur 2	1.95	4.05	0.58	1.97
SE on 3	2.09			
SE off 3	2.49			
SE Dur 3	0.40			
SE on 4	1.52	2.52	1.68	1.48

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

gastrocnemius lateral head

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

Boots	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	22.88	21.41	21.70	23.28	20.20	20.96	25.37	23.08	22.21	28.42	21.60	22.09
ave dur	22.88	21.41	21.70	23.28	20.20	20.96	25.37	23.08	22.21	28.42	21.60	22.09
ave on2	27.75	25.00	30.95	29.75	30.53	29.07			33.80		33.30	
ave off2	49.80	39.10	52.60	52.80	48.47	44.17			49.80		58.30	
ave dur2	22.10	14.10	21.65	23.00	23.90	15.13			15.95		25.00	
ave on3	88.59	85.48	87.48	85.83	83.54	83.17	85.83	87.36	85.45	85.41	81.63	81.62
ave dur3	25.14	21.30	22.28	24.94	21.22	21.88	26.40	23.42	22.90	28.80	22.70	21.88
ave on4	23.84	21.74	21.65	23.07	21.24	21.28	25.11	22.79	22.81	27.90	22.26	21.90
SE off	2.11	1.09	1.14	1.39	1.26	1.36	1.24	1.15	0.75	1.76	1.40	0.85
SE Dur	2.11	1.09	1.14	1.39	1.26	1.36	1.24	1.15	0.75	1.76	1.40	0.85
SE on2	1.67		0.02	1.59	2.18	0.35			0.22			
SE off2	5.23		1.48	4.56	3.14	1.19			3.09			
SE Dur2	3.58		1.50	2.95	4.30	1.53			3.29			
SE on3	2.19	1.58	1.29	1.71	1.30	1.67	1.63	1.60	1.64	2.21	1.60	1.34
SE dur 3	2.58	1.33	1.38	1.31	1.22	1.40	0.80	0.94	0.92	2.06	1.26	0.61
SE on4	2.05	1.20	1.19	1.56	0.99	1.13	1.20	0.88	0.71	1.76	0.97	0.50

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

key:

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gastrocnemius lateral head

Training shoes

	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	24.69	23.61	21.56	25.27	24.48	22.37	25.01	21.46	21.40	25.81	22.38	21.75
ave dur	24.69	23.61	21.56	25.27	24.48	22.37	25.01	21.46	21.40	25.81	22.38	21.75
ave on2	29.95			34.60	35.10	33.90	35.00	33.70				
ave off2	54.55			65.40	57.90	48.40	48.30	47.40				
ave dur2	24.60			30.80	22.80	17.70	13.30	13.70				
ave on3	86.51	87.44	86.34	83.50	85.80	83.50	84.69	87.16	84.95	83.00	85.10	84.16
ave dur3	25.18	23.28	21.68	23.48	24.33	22.25	26.63	21.53	22.23	27.95	22.55	21.35
ave on4	25.90	24.56	22.16	26.51	26.49	23.60	25.90	21.81	22.33	26.01	22.57	22.07
SE off	1.72	1.61	0.75	2.40	2.26	2.09	1.07	0.83	0.83	1.63	0.56	1.20
SE Dur	1.72	1.61	0.75	2.40	2.26	2.09	1.07	0.83	0.83	1.63	0.56	1.20
SE on2	0.60							2.40				
SE off2	0.96							4.38				
SE Dur2	0.36							1.98				
SE on3	1.59	1.88	1.66	1.35	1.99	1.32	0.51	1.79	1.76	1.76	1.38	1.66
SE dur 3	0.34	0.51	0.85	1.09	1.70	1.00	1.02	0.81	0.99	1.76	0.69	1.40
SE on4	1.64	1.73	0.67	2.60	2.15	2.22	0.84	0.65	0.75	1.61	0.57	1.05

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

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gastrocnemius lateral head

	Boots		Training shoes	
	jog straight	run straight	jog straight	run straight
ave off	21.16	21.19	23.53	21.83
ave dur	21.16	21.19	23.53	21.83
ave on2	41.40	32.68	31.68	33.90
ave off2	59.00	42.70	44.55	58.00
ave dur2	17.58	10.00	12.85	24.10
ave on3	85.12	85.08	87.68	81.38
ave dur3	22.16	20.73	23.16	20.78
ave on4	21.92	20.84	23.36	22.44
SE off	1.34	1.26	1.06	1.20
SE Dur	1.34	1.26	1.06	1.20
SE on2	3.85	0.66	1.33	
SE off2	8.54	0.47	0.87	
SE Dur2	4.71	1.12	0.47	
SE on3	1.21	2.08	1.63	1.35
SE dur 3	1.47	1.02	1.33	0.76
SE on4	1.24	0.78	1.19	1.26

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

gastocnemius medial head

Boots

	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
AVE OFF 1	23.44	22.83	21.67	26.48	21.82	20.74	23.87	23.13	22.49	26.13	21.72	22.15
AVE DUR	23.44	22.83	21.67	26.48	21.82	20.74	23.87	23.13	22.49	26.13	21.72	22.15
AVE ON 3	85.52	85.37	82.71	78.78	80.83	79.72	84.20	83.92	83.36	81.14	81.35	79.46
SE off	1.36	0.87	1.22	1.10	1.19	1.36	0.69	0.90	0.73	0.85	1.06	0.75
SE Dur	1.36	0.87	1.22	1.10	1.19	1.36	0.69	0.90	0.73	0.85	1.06	0.75
SE on3	1.28	1.29	1.48	4.44	2.12	1.58	1.12	1.11	0.82	3.91	1.90	1.76

Training shoes

	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
AVE OFF 1	24.71	24.65	21.60	26.72	22.66	23.41	24.67	22.16	21.38	23.98	21.23	21.52
AVE DUR	24.71	24.65	21.60	26.72	22.66	23.41	24.67	22.16	21.38	23.98	21.23	21.52
AVE ON 3	85.07	85.21	83.89	84.43	83.21	83.08	84.69	83.38	82.12	80.87	81.58	79.57
SE off	0.80	1.39	0.45	1.17	1.05	1.04	1.12	0.67	0.65	1.03	0.28	1.38
SE Dur	0.80	1.39	0.45	1.17	1.05	1.04	1.12	0.67	0.65	1.03	0.28	1.38
SE on3	1.94	1.47	1.10	1.40	1.01	1.32	1.34	0.74	1.47	1.17	1.74	1.24

	Boots		Training shoes	
	jog	run	jog	run
	straight	straight	straight	straight
AVE OFF 1	20.73	20.52	22.62	21.55
AVE DUR	20.73	20.52	22.62	21.55
AVE ON 3	83.36	81.83	85.20	79.91
SE off	0.78	1.26	1.03	1.22
SE Dur	0.78	1.26	1.03	1.22
SE on3	1.46	1.68	1.45	1.46

key:

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SE = standard error

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

Tensor fascia Latae

Boots	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	15.55	13.98	14.38	15.89	13.15	15.37	14.87	15.00	13.61	16.58	13.54	12.87
ave dur	15.55	13.98	14.38	15.89	13.15	15.37	14.87	15.00	13.61	16.58	13.54	12.87
ave on2	27.56	29.18	28.32	28.18	29.16	29.47	34.10	30.37	31.24	26.94	29.52	30.16
ave off2	58.25	59.19	55.70	63.25	59.87	61.60	55.85	55.62	57.52	61.93	57.84	60.14
ave dur2	30.68	29.99	27.40	35.06	30.72	32.13	21.75	24.96	26.92	34.97	29.26	29.98
ave on3	91.05	91.24	91.14	88.08	88.34	91.27	89.87	90.63	89.98	87.10	88.23	89.14
SE off	1.12	1.21	1.97	0.75	0.76	1.02	1.33	1.07	1.46	1.47	1.03	1.12
SE dur	1.12	1.21	1.97	0.75	0.76	1.02	1.33	1.07	1.46	1.47	1.03	1.12
SE on2	2.21	1.48	1.27	2.05	1.75	2.27	2.31	1.42	1.06	2.16	1.68	1.66
SE off2	2.69	1.78	2.18	1.85	2.12	3.02	2.29	0.83	1.41	2.05	1.80	1.37
SE dur2	2.00	1.66	2.24	1.92	2.18	2.02	1.98	2.01	1.10	2.27	1.46	1.86
SE on3	1.81	2.32	1.83	2.15	2.30	2.16	1.77	1.92	1.55	2.45	1.96	1.45

Training shoes	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
ave off	17.10	15.47	13.74	16.92	14.36	14.11	17.54	15.67	14.57	22.10	14.52	13.40
ave dur	17.10	15.47	13.74	16.92	14.36	14.11	17.54	15.67	14.57	22.10	14.52	13.40
ave on2	31.69	30.37	29.61	29.20	29.41	26.87	32.40	30.97	28.11	27.28	28.47	29.08
ave off2	59.51	59.32	58.98	63.60	59.77	60.53	58.76	58.91	56.14	57.16	58.74	56.75
ave dur2	27.84	28.66	29.38	34.42	30.36	33.67	26.37	27.96	26.53	29.87	30.29	27.70
ave on3	92.66	93.06	91.37	88.22	92.41	89.85	86.36	90.05	90.20	83.99	87.14	89.34
SE off	0.74	1.14	1.15	1.18	1.15	1.47	0.81	0.78	1.00	4.70	1.27	1.53
SE dur	0.74	1.14	1.15	1.18	1.15	1.47	0.81	0.78	1.00	4.70	1.27	1.53
SE on2	1.31	1.82	1.49	1.82	1.23	1.39	1.52	1.60	1.70	1.20	1.60	1.70
SE off2	1.66	2.03	0.93	2.33	0.96	2.73	1.88	1.64	1.60	1.59	0.70	1.72
SE dur2	1.83	1.54	1.80	3.36	0.99	2.51	2.25	1.34	2.42	1.63	1.46	2.11
SE on3	1.72	1.98	1.81	1.69	1.23	1.65	2.03	0.91	1.91	2.16	2.72	1.39

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

Tensor fascia Latae

	Boots		Training shoes	
	jog straight	run straight	jog straight	run straight
ave off	12.03	12.90	13.89	12.48
ave dur	12.03	12.90	13.89	12.48
ave on2	31.16	27.10	30.05	27.78
ave off2	55.53	57.03	56.59	54.91
ave dur2	25.18	29.94	26.54	28.63
ave on3	89.78	89.23	91.87	89.20
SE off	0.91	1.30	1.08	1.20
SE dur	0.91	1.30	1.08	1.20
SE on2	1.58	1.26	1.24	2.06
SE off2	1.89	1.82	1.41	1.27
SE dur2	1.96	2.23	2.01	2.17
SE on3	1.87	2.34	2.01	1.23

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

These temporal data represent mean values (with corresponding standard errors) across all subjects. Values are expressed as a % of stride cycle.

Gluteus Maximus

Boots

	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
AVE OFF	14.54	15.49	14.29	16.09	15.10	15.33	18.12	17.32	15.79	18.50	15.85	16.00
AVE DUR	14.54	15.49	14.29	16.09	15.10	15.33	18.12	17.32	15.79	18.50	15.85	16.00
AVE ON 2	72.82	73.48	72.94	69.90	70.55	66.09	71.01	73.57	70.61	69.45	70.82	67.17
SE off	1.34	1.78	2.01	1.79	1.79	1.93	1.28	1.15	1.49	1.76	1.65	1.43
SE Dur	1.34	1.78	2.01	1.79	1.79	1.93	1.28	1.15	1.49	1.76	1.65	1.43
SE on2	4.19	4.06	4.02	1.31	1.94	2.88	3.14	3.82	3.42	3.71	3.50	3.19

Training shoes

	Inside leg						Outside leg					
	jog			run			jog			run		
	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius	5m radius	10m radius	15m radius
AVE OFF	15.75	16.03	14.36	14.78	14.10	13.24	17.48	16.51	15.47	17.12	15.02	15.42
AVE DUR	15.72	16.03	14.36	14.78	14.10	13.24	17.48	16.51	15.47	17.12	15.02	15.42
AVE ON 2	78.37	78.02	74.53	72.44	74.47	71.24	75.01	77.75	76.02	68.76	70.29	71.12
SE off	1.29	0.79	0.87	1.48	0.87	0.94	1.04	0.63	1.24	1.36	1.64	1.69
SE Dur	1.30	0.79	0.87	1.48	0.87	0.94	1.04	0.63	1.24	1.36	1.64	1.69
SE on2	3.34	3.18	3.78	2.85	2.54	3.58	2.20	2.57	3.05	1.75	1.67	4.06

	boots		Training shoes	
	jog	run	jog	run
	straight	straight	straight	straight
AVE OFF	14.06	14.33	15.83	14.01
AVE DUR	14.06	14.33	15.83	14.01
AVE ON 2	75.37	69.29	76.42	73.04
SE off	1.14	1.03	0.54	0.91
SE Dur	1.14	1.03	0.54	0.91
SE on2	3.80	2.61	2.58	3.03

key:

off = cessation of activity
dur = duration of activity

on = restart of activity
SE = standard error

APPENDIX B

(Relevant to Chapter 4)

Foot contact time

These data show the mean and standard error of the foot contact time in seconds

Mean

	speed	heel time	toe time	footime	sf	sl	period
straight jog	4.43	0.12	0.19	0.31	1.44	3.09	0.70
straight run	5.41	0.11	0.18	0.29	1.54	3.53	0.65
15m jog in	4.39	0.11	0.19	0.30	1.44	3.05	0.70
15m run in	5.39	0.09	0.17	0.26	1.60	3.38	0.63
15m jog out	4.46	0.13	0.16	0.29	1.44	3.10	0.69
15m run out	5.54	0.13	0.16	0.28	1.60	3.47	0.63
10m jog in	4.50	0.11	0.19	0.30	1.42	3.18	0.71
10m run in	5.43	0.10	0.17	0.26	1.58	3.45	0.64
10m jog out	4.47	0.15	0.16	0.31	1.41	3.14	0.71
10m run out	5.21	0.13	0.16	0.30	1.56	3.46	0.64
5m jog in	4.47	0.12	0.17	0.29	1.44	3.10	0.69
5m run in	5.32	0.11	0.18	0.27	1.64	3.27	0.61
5m jog out	4.47	0.16	0.14	0.30	1.47	3.05	0.68
5m run out	5.37	0.15	0.16	0.32	1.70	3.17	0.59

Standard Error

	speed	heel time	toe time	footime	sf	sl	period
straight jog	0.05	0.01	0.01	0.01	0.02	0.05	0.01
straight run	0.08	0.01	0.01	0.02	0.03	0.05	0.01
15m jog in	0.03	0.01	0.01	0.01	0.02	0.05	0.01
15m run in	0.07	0.01	0.01	0.01	0.03	0.06	0.01
15m jog out	0.06	0.01	0.02	0.02	0.02	0.05	0.01
15m run out	0.08	0.01	0.02	0.02	0.03	0.10	0.01
10m jog in	0.03	0.01	0.01	0.01	0.02	0.05	0.01
10m run in	0.07	0.01	0.01	0.01	0.04	0.07	0.02
10m jog out	0.06	0.01	0.02	0.02	0.02	0.04	0.01
10m run out	0.11	0.01	0.03	0.02	0.03	0.06	0.01
5m jog in	0.04	0.01	0.01	0.01	0.03	0.06	0.01
5m run in	0.05	0.01	0.02	0.01	0.05	0.12	0.02
5m jog out	0.04	0.01	0.02	0.02	0.03	0.06	0.01
5m run out	0.10	0.01	0.03	0.03	0.06	0.12	0.02

key:

in = inside leg of curve
out = outside leg of curve

APPENDIX C

(Relevant to Chapter 5)

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike min = minimum displacement throughout cycle max = maximum displacement throughout cycle

min/max supp = minimum and maximum values during support

Inside leg - hip

	trial	min	max	heel st	min supp	max supp	toe off	heel st2
mean	straight	133.93	175.25	153.61	151.61	174.64	163.50	156.66
S.E.	jog	2.78	0.56	1.31	1.39	0.66	1.69	1.53
mean	10m	134.88	173.56	158.44	156.21	172.58	163.25	158.08
S.E.	jog	3.18	0.72	1.22	1.42	0.85	1.01	2.04
mean	7.5m	133.08	172.92	156.48	154.10	171.42	164.29	154.94
S.E.	jog	2.64	0.56	3.31	2.73	0.79	1.17	2.45
mean	5m	131.04	171.96	159.21	154.50	170.04	163.25	153.75
S.E.	jog	3.43	0.81	2.56	2.34	1.29	2.03	1.93

	trial	min	max	heel st	min supp	max supp	toe off	heel st2
mean	straight	130.98	174.69	150.79	149.96	174.02	163.40	154.81
S.E.	run	2.73	0.49	1.70	1.57	0.57	1.59	2.41
mean	10m	129.81	172.77	155.33	153.79	171.15	162.08	153.21
S.E.	run	2.10	0.59	3.20	2.92	1.02	1.52	1.93
mean	7.5m	126.04	171.13	152.54	151.42	170.31	160.98	151.94
S.E.	run	2.50	0.76	3.18	3.02	0.75	1.72	2.79
mean	5m	121.52	168.83	152.56	151.23	167.58	162.35	149.29
S.E.	run	1.61	1.01	3.29	2.91	0.68	1.58	2.24

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike min = minimum displacement throughout cycle max = maximum displacement throughout cycle

min/max supp = minimum and maximum values during support

Inside leg - knee

	trial	min	max	heel st	max supp	toe off	heel st2
mean	straight	61.38	173.63	162.06	142.57	157.53	167.50
S.E.	jog	4.22	0.84	1.13	0.99	1.70	1.29
mean	10m	63.69	173.94	164.56	145.94	154.46	168.38
S.E.	jog	4.51	0.73	1.67	1.07	2.18	1.53
mean	7.5m	62.50	172.04	163.96	146.94	156.00	166.58
S.E.	jog	3.11	1.32	1.40	0.97	1.82	1.35
mean	5m	63.38	171.79	162.92	149.38	154.75	167.92
S.E.	jog	2.35	1.28	2.39	2.14	2.16	1.73

	trial	min	max	heel st	max supp	toe off	heel st2
mean	straight	57.98	172.15	158.98	142.52	158.02	167.54
S.E.	run	3.04	1.36	2.72	2.38	1.91	1.37
mean	10m	57.35	171.81	163.23	146.52	152.88	167.33
S.E.	run	2.81	1.61	2.62	1.57	3.36	1.41
mean	7.5m	56.98	168.65	163.31	146.15	152.17	161.77
S.E.	run	2.84	2.26	2.69	1.70	2.80	2.43
mean	5m	57.83	166.15	161.10	149.79	148.40	160.71
S.E.	run	1.45	1.63	2.68	2.63	3.18	1.49

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike   min = minimum displacement throughout cycle   max = maximum displacement throughout cycle

min/max supp = minimum and maximum values during support

Inside leg - ankle

	trial	min	max	heel st	max supp	toe off	heel st2
mean	straight	89.13	134.73	107.62	89.07	131.37	107.81
S.E.	jog	1.44	1.70	1.18	1.41	2.00	2.19
mean	10m	88.98	130.75	110.50	88.98	129.17	108.98
S.E.	jog	1.43	1.39	1.76	1.40	1.40	2.20
mean	7.5m	87.50	130.29	110.29	87.54	127.96	109.25
S.E.	jog	1.52	3.03	1.77	1.50	2.92	1.45
mean	5m	88.75	128.79	112.92	88.71	126.50	112.50
S.E.	jog	1.39	3.66	2.18	1.45	3.16	4.02

	trial	min	max	heel st	max supp	toe off	heel st2
mean	straight	86.23	134.83	108.17	86.19	132.25	110.44
S.E.	run	1.61	2.41	1.26	1.60	2.22	2.22
mean	10m	87.67	131.83	110.52	87.67	128.17	110.29
S.E.	run	1.45	1.67	2.41	1.44	1.53	2.38
mean	7.5m	85.69	131.15	113.15	85.75	127.48	110.94
S.E.	run	1.09	3.01	2.48	1.19	3.02	1.98
mean	5m	88.27	127.50	111.40	88.15	121.67	111.35
S.E.	run	2.18	4.05	3.23	2.14	4.21	3.14

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike   min = minimum displacement throughout cycle   max = maximum displacement throughout cycle

min/max supp = minimum and maximum values during support

Outside leg - hip

	trial	min	max	heel st	min supp	max supp	toe off
mean	straight	133.36	174.86	157.08	155.69	172.61	159.47
S.E.	jog	2.39	0.74	1.71	1.67	0.62	1.40
mean	10m	135.67	175.65	158.35	156.17	174.02	161.98
S.E.	jog	3.11	0.42	1.40	0.68	0.62	1.32
mean	7.5m	133.81	175.15	157.10	155.63	173.96	162.21
S.E.	jog	2.15	0.33	1.15	1.23	0.66	1.65
mean	5m	132.25	174.96	156.71	154.25	172.96	163.67
S.E.	jog	2.11	0.57	1.74	1.52	0.72	2.17

	trial	min	max	heel st	min supp	max supp	toe off
mean	straight	129.17	174.60	154.48	153.48	172.40	157.73
S.E.	run	2.49	0.61	1.37	1.37	0.68	2.32
mean	10m	128.88	174.88	154.21	152.94	173.06	162.56
S.E.	run	2.00	0.78	2.48	1.96	0.66	2.17
mean	7.5m	128.19	174.40	151.65	150.69	172.56	162.38
S.E.	run	1.92	1.12	1.73	1.70	1.04	2.08
mean	5m	128.21	174.67	152.96	151.06	173.17	164.48
S.E.	run	1.97	0.96	2.07	1.96	1.11	2.17

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike
min = minimum displacement throughout cycle
max = maximum displacement throughout cycle
min/max supp = minimum and maximum values during support

Outside leg - knee

	trial	min	max	heel st	max supp	toe off
mean	straight	61.22	171.52	165.89	146.21	159.18
S.E.	jog	2.28	0.93	0.50	1.03	1.49
mean	10m	63.81	171.13	163.19	141.60	159.27
S.E.	jog	4.46	1.46	2.28	1.10	2.49
mean	7.5m	62.42	169.94	161.17	140.04	159.83
S.E.	jog	2.77	1.05	1.15	1.57	2.11
mean	5m	64.00	168.96	161.00	141.75	158.63
S.E.	jog	1.87	1.12	1.63	1.79	1.90

trial	min	max	heel st	max supp	toe off
straight	56.33	169.10	163.02	144.15	161.31
run	1.96	1.62	1.60	1.89	1.39
10m	56.54	166.67	158.38	141.67	154.31
run	3.02	1.81	2.86	1.90	3.95
7.5m	57.13	165.29	155.54	138.08	157.25
run	2.30	0.94	1.83	1.49	1.70
5m	60.48	165.92	155.21	139.69	158.29
run	2.23	1.25	1.59	1.59	2.27

The data below represents angular displacement values during straight and curvilinear motion when jogging and running (degrees)
Mean values with standard errors are presented at key points of the stride cycle from right heel strike to right heel strike

heel st = heel strike   min = minimum displacement throughout cycle   max = maximum displacement throughout cycle

min/max supp = minimum and maximum values during support

outside leg - ankle

	trial	min	max	heel st	max supp	toe off	trial	min	max	heel st	max supp	toe off
mean	straight	84.79	135.11	104.80	84.89	127.88	straight	85.29	139.00	106.02	85.21	127.31
S.E.	jog	1.28	1.34	1.82	1.20	2.33	run	1.77	2.34	2.34	1.78	2.50
mean	10m	84.77	135.15	106.04	84.88	130.88	10m	89.83	135.83	107.67	89.73	129.67
S.E.	jog	1.63	2.47	1.85	1.68	1.52	run	2.05	2.54	1.76	2.05	2.09
mean	7.5m	85.71	134.10	106.15	85.73	128.31	7.5m	88.21	135.60	107.25	88.31	128.13
S.E.	jog	1.14	3.18	1.25	1.03	2.57	run	1.04	3.35	0.89	0.99	2.26
mean	5m	88.67	130.83	107.08	88.63	128.63	5m	92.35	134.71	107.50	92.56	128.25
S.E.	jog	1.53	3.14	2.59	1.54	2.46	run	3.06	3.45	2.17	3.23	3.71

APPENDIX D

(Relevant to Chapter 6)

Columns shown

The following data are mean values for all subjects with associated standard errors
A key for the selected variables is shown below:-

contact	Foot contact time (seconds)
air time	Ballsitic air time between foot contacts (seconds)
time 1st max	Time to first vertical impact peak (seconds)
1st max	First vertical impact peak (BW)
time 1st min	Time to first vertical impact trough (seconds)
1st min	First vertical impact trough (BW)
free mom max	Anticlockwise free moment (Nm)
free mom min	Clockwise free moment (Nm)
2nd max	Vertical drive-off peak (BW)
average force	Average vertical ground reaction force (BW)
Fz impulse	Vertical impulse (Ns)
total force max	Maximum total force in all three axes (BW)
total force ave	Average total force in all three axes (BW)
Fx max	Maximum mediolateral force (BW)
Fx min	Minimum mediolateral force (BW)
Fx impulse	Mediolateral impulse (Ns)
Fy min	Maximum braking force (BW)
braking impulse	Braking impulse (Ns)
time to 0	Time to beginning of propulsion (seconds)
Fy max	Maximum propulsion force (BW)
propulse impulse	Propulsion impulse (Ns)

out = outside leg of curve
in = inside leg of curve

Row order of TRIALS shown in left hand column

Straight jog	shoes
	boots
Straight run	shoes
	boots
5m jog	shoes
	boots
5m run	shoes
	boots

2 force plates

	mass	contact outside	contact inside	air time	time 1st max out	time 1st max in	1st max outside	1st max inside	time 1st min out	time 1st min in	1st min outside	1st min inside
shoe straight jog												
mean	**808.22**	**0.22**	**0.23**	**0.16**	**0.13**	**0.11**	**2.27**	**2.37**	**0.19**	**0.17**	**1.95**	**1.85**
se	43.89	0.01	0.01	0.01	0.01	0.01	0.24	0.15	0.01	0.01	0.09	0.17
boot straight jog												
mean	**808.33**	**0.22**	**0.23**	**0.16**	**0.12**	**0.10**	**2.50**	**2.58**	**0.18**	**0.16**	**1.68**	**1.62**
se	43.83	0.01	0.01	0.01	0.01	0.01	0.27	0.10	0.01	0.01	0.07	0.14
shoe straight run												
mean	**813.33**	**0.19**	**0.20**	**0.14**	**0.13**	**0.12**	**2.50**	**2.49**	**0.19**	**0.18**	**1.98**	**1.85**
se	41.16	0.00	0.00	0.00	0.01	0.01	0.18	0.11	0.01	0.01	0.10	0.15
boot straight run												
mean	**808.33**	**0.19**	**0.20**	**0.15**	**0.12**	**0.15**	**2.73**	**2.68**	**0.18**	**0.17**	**1.65**	**1.64**
se	43.83	0.01	0.00	0.00	0.01	0.05	0.14	0.11	0.00	0.01	0.17	0.13
shoe 5m jog												
mean	**804.33**	**0.21**	**0.24**	**0.09**	**0.16**	**0.12**	**2.10**	**1.63**	**0.21**	**0.17**	**1.86**	**1.43**
se	37.78	0.01	0.01	0.01	0.01	0.01	0.16	0.12	0.01	0.01	0.14	0.06
boot 5m jog												
mean	**804.33**	**0.22**	**0.24**	**0.09**	**0.13**	**0.08**	**2.00**	**1.74**	**0.18**	**0.13**	**1.61**	**1.37**
se	37.78	0.01	0.01	0.01	0.01	0.01	0.13	0.14	0.01	0.01	0.05	0.07
shoe 5m run												
mean	**801.50**	**0.20**	**0.22**	**0.07**	**0.13**	**0.14**	**2.30**	**1.50**	**0.19**	**0.15**	**1.91**	**1.29**
se	39.55	0.01	0.01	0.01	0.01	0.03	0.07	0.14	0.02	0.01	0.11	0.12
boot 5m run												
mean	**802.50**	**0.20**	**0.21**	**0.07**	**0.11**	**0.09**	**2.03**	**1.84**	**0.15**	**0.12**	**1.67**	**1.33**
se	39.82	0.01	0.01	0.01	0.01	0.01	0.12	0.22	0.02	0.01	0.08	0.09

2 force plates

	Free mom max out	Free mom max in	Free mom min out	Free mom min in	2nd max outside	2nd max inside	average force out	average force in	Fz impulse outside	Fz impulse inside
shoe straight jog										
mean	**27.89**	**17.55**	**-21.55**	**-19.57**	**2.95**	**2.70**	**1.71**	**1.64**	**306.28**	**305.22**
se	6.95	3.01	4.09	5.63	0.10	0.09	0.07	0.07	20.11	18.22
boot straight jog										
mean	**27.67**	**17.64**	**-31.02**	**-20.07**	**2.92**	**2.78**	**1.68**	**1.64**	**301.44**	**303.78**
se	5.48	3.37	8.85	3.07	0.18	0.11	0.05	0.06	19.23	16.48
shoe straight run										
mean	**14.52**	**17.02**	**-46.34**	**-26.42**	**2.94**	**2.75**	**1.77**	**1.65**	**276.94**	**270.00**
se	2.07	3.71	23.07	6.59	0.10	0.10	0.05	0.04	16.05	15.13
boot straight run										
mean	**17.84**	**21.46**	**-20.00**	**-19.22**	**2.99**	**2.82**	**1.77**	**1.68**	**275.22**	**274.94**
se	3.97	3.13	6.96	7.39	0.12	0.09	0.06	0.06	14.43	12.45
shoe 5m jog										
mean	**39.46**	**26.29**	**-76.29**	**-59.04**	**2.57**	**2.29**	**1.52**	**1.32**	**262.83**	**256.28**
se	16.90	11.54	11.95	8.70	0.12	0.06	0.06	0.03	15.20	9.23
boot 5m jog										
mean	**50.05**	**29.29**	**-43.14**	**-59.29**	**2.51**	**2.32**	**1.51**	**1.37**	**265.22**	**261.89**
se	13.48	15.13	6.21	9.94	0.10	0.10	0.04	0.05	14.51	8.06
shoe 5m run										
mean	**63.90**	**25.27**	**-61.44**	**-53.09**	**2.46**	**2.03**	**1.52**	**1.27**	**238.61**	**222.60**
se	17.89	10.15	16.04	3.73	0.07	0.10	0.05	0.06	7.65	8.97
boot 5m run										
mean	**59.49**	**21.45**	**-65.50**	**-64.22**	**2.49**	**2.20**	**1.51**	**1.30**	**243.92**	**222.56**
se	14.57	8.45	16.86	11.85	0.07	0.19	0.05	0.07	11.19	13.02

	total Force max out	total force max in	total Force ave outside	total Force ave inside	Fx max outside	2 force plates Fx max inside	Fx min outside	Fx min inside	Fx ave outside	Fx ave inside	Fx impulse outside	Fx impulse inside
shoe straight jog												
mean	**3.03**	**2.76**	**1.74**	**1.67**	**0.15**	**0.16**	**-0.14**	**-0.10**	**0.00**	**0.01**	**0.16**	**2.18**
se	0.13	0.11	0.07	0.07	0.05	0.03	0.02	0.02	0.02	0.01	2.97	1.39
boot straight jog												
mean	**3.07**	**2.84**	**1.71**	**1.67**	**0.16**	**0.19**	**-0.14**	**-0.11**	**0.00**	**0.01**	**0.48**	**2.20**
se	0.15	0.09	0.06	0.06	0.05	0.03	0.02	0.01	0.02	0.01	3.11	1.22
shoe straight run												
mean	**2.97**	**2.78**	**1.80**	**1.69**	**0.11**	**0.21**	**-0.19**	**-0.12**	**-0.02**	**0.01**	**-2.68**	**1.93**
se	0.11	0.10	0.06	0.04	0.02	0.02	0.03	0.03	0.02	0.01	2.39	2.32
boot straight run												
mean	**3.01**	**2.86**	**1.81**	**1.72**	**0.11**	**0.20**	**-0.17**	**-0.12**	**-0.01**	**-0.02**	**-1.55**	**-0.14**
se	0.13	0.10	0.06	0.06	0.02	0.01	0.02	0.02	0.02	0.04	2.32	2.23
shoe 5m jog												
mean	**2.83**	**2.36**	**1.65**	**1.38**	**1.09**	**0.55**	**-0.03**	**-0.03**	**0.57**	**0.31**	**98.83**	**59.39**
se	0.12	0.08	0.07	0.04	0.07	0.06	0.01	0.01	0.04	0.04	8.64	6.67
boot 5m jog												
mean	**2.72**	**2.41**	**1.63**	**1.45**	**1.02**	**0.65**	**-0.04**	**-0.02**	**0.54**	**0.37**	**94.59**	**69.71**
se	0.08	0.11	0.04	0.06	0.03	0.06	0.01	0.01	0.02	0.04	7.91	4.08
shoe 5m run												
mean	**2.76**	**2.13**	**1.67**	**1.36**	**1.20**	**0.63**	**-0.04**	**-0.02**	**0.61**	**-0.37**	**92.70**	**64.18**
se	0.08	0.13	0.06	0.07	0.09	0.07	0.01	0.01	0.04	0.05	5.36	6.10
boot 5m run												
mean	**2.74**	**2.33**	**1.68**	**1.40**	**1.15**	**0.75**	**-0.05**	**0.04**	**0.63**	**0.43**	**102.81**	**71.79**
se	0.09	0.21	0.06	0.08	0.07	0.08	0.01	0.06	0.05	0.05	9.17	7.01

2 force plates

	Fy min outside	Fy min inside	braking impulse ou	braking impulse in	time to 0 outside	time to 0 inside	Fy max outside	Fy max inside	propulse impulse ou	propulse impulse in
shoe straight jog										
mean	**-0.63**	**-0.58**	**-26.16**	**-27.16**	**0.51**	**0.53**	**0.36**	**0.37**	**17.04**	**17.86**
se	0.13	0.03	5.75	1.55	0.02	0.01	0.03	0.02	2.24	1.82
boot straight jog										
mean	**-0.59**	**-0.59**	**-24.68**	**-26.24**	**0.49**	**0.45**	**0.41**	**0.37**	**21.03**	**18.80**
se	0.11	0.05	5.57	1.23	0.02	0.06	0.03	0.03	1.01	1.51
shoe straight run										
mean	**-0.63**	**-0.68**	**-22.65**	**-20.25**	**0.50**	**0.51**	**0.45**	**0.44**	**19.36**	**19.49**
se	0.09	0.06	3.82	2.44	0.02	0.01	0.02	0.02	2.14	1.29
boot straight run										
mean	**-0.60**	**-0.73**	**-21.41**	**-26.49**	**0.48**	**0.50**	**0.48**	**0.44**	**20.91**	**19.20**
se	0.08	0.06	3.29	1.13	0.01	0.01	0.02	0.02	2.29	2.44
shoe 5m jog										
mean	**-0.44**	**-0.44**	**-16.06**	**-27.34**	**0.48**	**0.55**	**0.41**	**0.23**	**20.04**	**9.15**
se	0.06	0.05	4.39	2.73	0.03	0.02	0.03	0.02	1.01	2.86
boot 5m jog										
mean	**-0.44**	**-0.53**	**-17.90**	**-30.15**	**0.45**	**0.54**	**0.39**	**0.21**	**19.19**	**8.92**
se	0.07	0.03	5.08	1.96	0.02	0.01	0.03	0.02	1.30	2.34
shoe 5m run										
mean	**-0.43**	**-0.44**	**-13.25**	**-27.27**	**0.47**	**0.55**	**0.45**	**0.25**	**20.15**	**9.78**
se	0.04	0.06	2.18	2.14	0.02	0.03	0.04	0.02	1.30	1.17
boot 5m run										
mean	**-0.44**	**-0.58**	**-15.22**	**-28.93**	**0.48**	**0.50**	**0.45**	**0.22**	**21.25**	**10.17**
se	0.04	0.07	2.65	2.65	0.03	0.06	0.05	0.02	2.36	1.25

APPENDIX E

(Relevant to Chapter 5)

Appendix E

Table 5.7 Mean angular displacement (degrees) of the hip during straight and curvilinear motion during jogging.

Joint	Trial	Min	Max	Range	1st Heel strike	Min supp.	Max supp.	Toe-off	2nd Heel strike
inside leg hip	st jog	133.9	175.2	41.3	153.6	151.6	174.6	163.5	156.7
inside leg hip	5m jog	131.0	172.0	41.0	159.2	154.5	170.0	163.3	153.8
outside leg hip	st jog	133.4	174.9	41.5	157.1	155.7	172.6	159.5	-
outside leg hip	5m jog	132.3	175.0	42.7	156.7	154.3	173.0	163.7	-

Table 5.8 Mean angular displacement values at the knee during straight and curvilinear motion during jogging.

Joint	Trial	Min	Max	Range	Heel strike	Max support	Toe-off	Heel strike 2
inside leg knee	st jog	61.4	173.6	112.2	162.0	142.6	157.5	167.5
inside leg knee	5m jog	63.4	171.8	108.4	162.9	149.4	154.8	167.9
outside leg knee	st jog	61.2	171.5	110.3	165.9	146.2	159.2	-
outside leg knee	5m jog	64.0	169.0	105.0	161.0	141.8	158.6	-

Table 5.9 Mean angular displacement values at the ankle during straight and curvilinear motion during jogging.

Joint	Trial	Min	Max	Range	Heel strike	Max support	Toe-off	Heel strike 2
inside ankle	st jog	89.1	134.7	45.6	107.6	89.1	131.4	107.8
inside ankle	5m jog	88.8	128.8	40.0	112.9	88.7	126.5	112.5
outside ankle	st jog	84.8	135.1	50.3	104.8	84.9	127.9	-
outside ankle	5m jog	88.7	130.8	42.1	107.1	88.6	128.6	-

APPENDIX F

(Relevant to Chapter 7)

Comparison of six-studded soccer boot with moulded sole boot during shooting, Drag-back and Cruyff turns
The following data are mean values for all subjects with associated standard errors
A key for the selected variables is shown below:-

Fz max	First vertical impact peak (BW)
mz max	Anticlockwise free moment (BWNm)
mz min	Clockwise free moment (BWNm)
Fx max	Maximum mediolateral force (BW)
Fx min	Minimum mediolateral force (BW)
Fy min	Maximum braking force (BW)
Fy max	Maximum propulsion force (BW)
tz min	Clockwise vertical torque (BWNm)
tz max	Anticlockwise vertical torque (BWNm)
fric ave	Average friction
fric max	Maximum friction

For the drag-back turn, some subjects performed with two discrete foot contacts, which are labelled 1 and 2.
For those subject using only one foot contact, just the variable name is listed

Drag back turn

VARIABLE		fz max	fz max contact 1	fz max contact 2	fy min	fy min 1	fy min 2	fy max	fy max 1	fy max 2
Standard six-stud	**mean**	2.16	2.28	1.78	-1.12	-1.02	-0.67	0.06	0.09	0.07
	S.E.	0.10	0.07	0.07	0.12	0.08	0.10	0.01	0.02	0.01
Moulded Traxion sole	**mean**	2.05	2.13	1.88	-1.01	-0.99	-0.82	0.09	0.05	0.10
	S.E.	0.09	0.08	0.05	0.07	0.05	0.03	0.02	0.01	0.01

VARIABLE		fx min	fx min 1	fx min 2	fx max	fx max 1	fx max 2	mz min	mz max	mz min 1	mz min 2
Standard six-stud	**mean**	-0.20	-0.13	-0.31	0.17	0.14	0.07	-0.40	0.49	-0.53	-0.41
	S.E.	0.01	0.02	0.12	0.05	0.03	0.01	0.07	0.14	0.06	0.10
Moulded Traxion sole	**mean**	-0.17	-0.13	-0.13	0.14	0.11	0.07	-0.36	0.55	-0.52	-0.28
	S.E.	0.02	0.02	0.02	0.03	0.03	0.01	0.05	0.08	0.06	0.07

VARIABLE		mz max 1	mz max 2	tz min	tz max	tz min 1	tz min 2	tz max 1	tz max 2
Standard six-stud	**mean**	0.12	0.33	-0.42	0.17	-0.28	-0.16	0.08	0.12
	S.E.	0.03	0.06	0.04	0.03	0.05	0.03	0.01	0.02
Moulded Traxion sole	**mean**	0.38	0.21	-0.35	0.12	-0.33	-0.16	0.08	0.10
	S.E.	0.11	0.03	0.05	0.01	0.05	0.04	0.01	0.01

VARIABLE		friction av	fric max	fric ave 1	fric ave 2	fric max 1	fric max 2
Standard six-stud	**mean**	0.53	3.51	0.52	0.46	2.52	2.88
	S.E.	0.03	0.56	0.03	0.03	0.51	0.29
Moulded Traxion sole	**mean**	0.50	4.00	0.50	0.58	1.65	4.75
	S.E.	0.02	0.53	0.02	0.08	0.24	0.47

Cruyff turn

VARIABLE		fz max	fy min	fy max	fx min	fx max	mz min	mz max	tz min	tz max	fr ave	fr max
Standard six-stud	**mean**	2.31	-1.03	0.05	-0.31	0.27	-0.56	0.34	-0.15	0.12	0.57	3.14
	S.E.	0.15	0.12	0.02	0.04	0.04	0.11	0.08	0.05	0.05	0.05	0.41
Moulded Traxion sole	**mean**	2.05	-0.94	0.05	-0.28	0.22	-0.45	0.32	-0.18	0.10	0.59	5.25
	S.E.	0.10	0.06	0.01	0.04	0.03	0.07	0.10	0.03	0.02	0.02	0.72

Shot

VARIABLE		fz max	fy min	fy max	fx min	fx max	mz min	mz max	tz min	tz max	fr ave	fr max
Standard six-stud	**mean**	3.95	-1.27	0.10	-0.04	0.80	-0.82	0.87	-0.16	0.11	0.37	3.31
	S.E.	0.14	0.11	0.03	0.01	0.06	0.19	0.13	0.03	0.01	0.01	0.41
Moulded Traxion sole	**mean**	3.74	-1.22	0.20	-0.06	0.67	-0.63	0.77	-0.18	0.09	0.39	4.68
	S.E.	0.10	0.10	0.06	0.01	0.06	0.15	0.16	0.03	0.02	0.02	0.52

APPENDIX G

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INFORMED CONSENT FORM

The project will investigate muscle activity generated in the lower extremity during soccer specific actions to provide an insight into muscle use during soccer.

The procedure for the study requires each subject to have single use disposable surface electrodes attached to the skin over six muscles of the right leg. Before attaching the electrodes the skin is cleaned and lightly scraped with a single subject disposable rasp made of a Velcro line material before a jelly is applied for a few minutes. The electrodes are attached to a medical grade (British Standard) device. This sends the muscle signals using radio waves to a computer where the muscle signals are recorded. Note: This device only records your own muscle signals and is powered by a 9 volt battery. It uses a protection system (British Standard) which only allows your muscle signals to pass to the device but nothing to pass from the device to your skin.

Each subject will complete a number of soccer specific actions whilst wearing soccer boots, and also wearing flat training shoes. The actions are straight jog and run, followed by curvilinear jog and run around circles of 15m, 10m, and 5m radius. Three types of soccer turns will also be performed.

During the movement the subjects will be recorded on video tape, the custody of which shall remain with the experimenters who will have sole access to it. The video recording will serve as a record of each trial and may also help identify key events. Upon completion of the study video material shall be destroyed.

Each subject is free to withdraw consent and stop taking part in the study at any time. This withdrawal will be without prejudice and no disadvantage will arise if you decide to no longer participate.

All data collected will be used in absolute confidentiality.

I (full name and date)

understand what is involved in the test and give my consent to participate in the test(s) explained to me.

Subject's signature

Experimenter's signature

Supervisors signature

APPENDIX H

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A Registered Charity

INFORMED CONSENT FORM

The project will investigate foot contact time during curved runing patterns to provide an insight into the way we move during soccer.

The procedure for the study requires each subject to have footpads located in a foam insole within the right soccer boot. The footpads are attached to a medical grade (British Standard) device. This sends the signals using radio waves to a computer where the foot cantact signals are recorded. Note: This device only records your own foot contact signals and is powered by a 9 volt battery. It uses a protection system (British Standard) which only allows signals to pass to the device but nothing to pass from the device to your boot.

Each subject will complete a number of curved running patterns whilst wearing soccer boots. The actions are straight jog and run, followed by curvilinear jog and run around circles of 15m, 10m, and 5m radius.

During the movement the subjects will be recorded on video tape, the custody of which shall remain with the experimenters who will have sole access to it. The video recording will serve as a record of each trial and may also help identify key events. Upon completion of the study video material shall be destroyed.

Each subject is free to withdraw consent and stop taking part in the study at any time. This withdrawal will be without prejudice and no disadvantage will arise if you decide to no longer participate.

All data collected will be used in absolute confidentiality.

I (full name and date)

understand what is involved in the test and give my consent to participate in the test(s) explained to me.

Subject's signature

Experimenter's signature

Supervisors signature

APPENDIX I

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A Registered Charity

INFORMED CONSENT FORM

The project will investigate patterns of body movement in the lower extremity during curved and straight running actions to provide an insight into how curved running actions are achieved in soccer.

Each subject will complete the running actions whilst wearing soccer boots. The actions are straight jog and run, followed by curvilinear jog and run around circles of 10 m, 7.5 m, and 5 m radii. Movement speed will be monitored by infrared timing lights (Cla-Win timer, University College Chichester) and trials must be performed at 4.5 m/s for the jog, and 5.5 m/s for the run.

During the movement the subjects will be recorded on videotape, the custody of which shall remain with the experimenters who will have sole access to it. The video recording will be used to investigate lower body kinematics. Upon completion of the study video material shall be destroyed.

Each subject is free to withdraw consent and stop taking part in the study at any time. This withdrawal will be without prejudice and no disadvantage will arise if you decide to no longer participate.

All data collected will be used in absolute confidentiality.

I (full name and date)

understand what is involved in the test and give my consent to participate in the test(s) explained to me.

Subject's signature

Experimenter's signature

Supervisors signature

APPENDIX J

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INFORMED CONSENT FORM

The project will investigate muscle activity generated in the lower extremity during curved and straight running actions to provide an insight into muscle use during soccer. Measures of ground reaction force will also be taken using a force platform. The force platform will be lodged securely in the ground and enables the measurement of the force with which the subject contacts the ground.

The procedure for the study requires each subject to have single use disposable surface electrodes attached to the skin over three muscles of the right leg. Before attaching the electrodes the skin is cleaned and lightly scraped with a single subject disposable rasp made of a Velcro line material before a jelly is applied for a few minutes. The electrodes are attached to a medical grade (British Standard) device. This sends the muscle signals using radio waves to a computer where the muscle signals are recorded. Note: This device only records your own muscle signals and is powered by a 9 volt battery. It uses a protection system (British Standard) which only allows your muscle signals to pass to the device but nothing to pass from the device to your skin.

Each subject will complete the running actions whilst wearing soccer boots, and also wearing flat training shoes. The actions are straight jog and run, followed by curvilinear jog and run around a circle of 5m radius.

During the movement the subjects will be recorded on video tape, the custody of which shall remain with the experimenters who will have sole access to it. The video recording will be used to investigate lower body kinematics. Upon completion of the study video material shall be destroyed.

Each subject is free to withdraw consent and stop taking part in the study at any time. This withdrawal will be without prejudice and no disadvantage will arise if you decide to no longer participate.

All data collected will be used in absolute confidentiality.

I (full name and date)

understand what is involved in the test and give my consent to participate in the test(s) explained to me.

Subject's signature

Supervisors signature

Experimenter's signature

APPENDIX K

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INFORMED CONSENT FORM

A Registered Charity

The project will investigate ground reaction force measures during soccer specific actions.

The procedure for the study will require each subject to perform the movements on a Force measuring platform inserted into a natural grass surface. The equipment will measure the forces created between the player and the ground as he turns or shoots. Each movement will be performed six times.

Each subject will complete a number of soccer specific actions whilst wearing two styles of soccer boots. The actions are two types of soccer turns , a change of direction of 90 degrees and a shot will also be performed.

During the movement the subjects will be recorded on video tape, the custody of which shall remain with the experimenters who will have sole access to it. The video recording will serve as a record of each trial and may also help identify key events. Upon completion of the study video material shall be destroyed.

Each subject is free to withdraw consent and stop taking part in the study at any time. This withdrawal will be without prejudice and no disadvantage will arise if you decide to no longer participate.

All data collected will be used in absolute confidentiality.

I (full name and date)

understand what is involved in the test and give my consent to participate in the test(s) explained to me.

Subject's signature

Experimenter's signature

Supervisors signature