AN INVESTIGATION OF THE RELATIONSHIP BETWEEN ANTERIOR KNEE PAIN AND LOWER LIMB BIOMECHANICS

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A thesis submitted in partial fulfilment of the requirements of the Robert Gordon University for the degree of Master of Research

January 2011

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ABSTRACT

The thesis presents an investigation of the relationship between anterior knee pain and lower limb biomechanics in symptomatic and asymptomatic individuals during walking.

Although foot orthotic devices are often prescribed to realign lower extremity mechanics, there is conflicting evidence to support this. A quantitative study explored the dynamics of walking barefoot, shod and with orthotic devices using the 3D Kinematic Motion Analysis System, Vicon 370. Sample sizes of 30 control subjects and 30 subjects with anterior knee pain were recruited. Kinematic gait parameters and temporal-spatial gait parameters were compared between the two groups and a thorough static clinical examination was provided.
The relationship of alignment and knee pain was also examined using the Positional Upright MRI Scanner. A group of 15 subjects with anterior knee pain and a group of 5 asymptomatic subjects were examined radiographically barefoot, shod and with orthotic devices.

An increase in Q-angle was shown in the knee pain group with an added angle increase in the females in both groups. This could signify that a higher Q-angle may influence biomechanics of the knee joint by creating an abnormally increased valgus angle. Evidence was found that there was a strong relationship between eversion and inversion of the foot with internal and external tibial rotation of the leg. This coupling relationship may lend insight into the behavior of the lower leg and anterior knee pain whilst walking. Small changes were noted when wearing the orthotic devices which may be advantageous biomechanically and they were also of benefit in correcting pronatory changes or movements more distal to the foot. The results of the MRI study were disappointing but many limitations were shown to support the results.

Upon finishing this thesis, it is apparent that documented evidence of any relationship between anterior knee pain and lower limb biomechanics is imperative to clinicians in order to aid in treatment plans of patients but also in preventative treatment. Further research is required to enhance the clinicians' knowledge and understanding of the foot and leg dynamically.

Chapter 1 – Introduction

1.1. Introduction

This chapter provides an introduction and justification to the origins and objectives of the thesis.

The aim of this thesis is to investigate any association between lower limb biomechanics and anterior knee pain in symptomatic and asymptomatic subjects during gait. There have been assumptions that altered foot mechanics as a primary problem can lead to secondary knee pain but the evidence in terms of objective analysis of lower limb and in particular foot biomechanics is limited. Some studies have suggested that differences in relative muscle forces exerted on the patella between the vastus medialis muscle and the vastus lateralis muscle contribute towards lateral patellar tracking and malalignment which have been linked to knee pain (Powers et al 1996; Cutbill et al 1997). Studies investigating foot function during gait have typically approximated foot pronation and supination using calcaneal eversion and inversion, as this component is the simplest to measure (Edington et al 1990). Deviation from the normal gait pattern, described as either excessive foot pronation or supination or calcaneal eversion and inversion depending on definitions and measurement technique, have been implicated as being contributory to a number of musculoskeletal disorders of the lower extremity.

Lower limb alignment was also investigated to establish any changes in barefoot, shod and orthotic conditions using a Motion Analysis System and a Positional MRI scanner. There are various theories from a review of literature whether foot orthotics affect lower limb biomechanics. It was believed that wearing shoes with and without the addition of foot orthoses could alter the kinematics of the lower limb therefore measuring motion at the rearfoot was paramount. The rearfoot (subtalar joint) is very difficult to examine clinically due to the fact it is structurally and functionally dependent on other joints forming the ankle joint and midtarsal complex. Due to this complexity, the subtalar joint presents many problems in the accurate measurement of range of motion. Previous literature does exist which has also investigated 2D and 3D motion at the rearfoot. Foot models have been devised for use with motion capture which enables 3D data to be collected at the subtalar joint. The foot model used in this study was slightly different to previous ones and this will be discussed in chapter 4 section 4.4. Another specific element of the new foot model involved was being able to measure navicular height both statically and dynamically. According to previous research, the measure of navicular height is a representation of pronation and supination but when the model was being developed, there had been no evidence in the research literature at the time on testing dynamic navicular height. One of the aims was to identify relationships between clinical measurements of lower limb biomechanics versus measurements taken dynamically using motion capture and positional MRI. In order to do this the reliability of clinical measurements, marker placement for use with motion capture and MRI measurements were tested in several pilot studies involving small samples of subjects. The repeatability, reliability and accuracy of the measurements were explored which will give confidence with the interpretation of the final results. The overall objective of this research was to gain a greater understanding of how the foot and leg work together and if there are any associations between the biomechanics of the lower leg and anterior knee pain during walking.

The findings of this study may be relevant for clinical practice and research in the following ways

a. Influence evidence based practice by presenting clinicians with data-supported clinical recommendations and guidelines, concerning the potential effectiveness for the use of functional foot orthoses in lower extremity injuries.
b. By providing a normative database of kinematic data using the control groups data which will enable clinicians to compare any symptomatic foot and leg kinematic data.
c. Development of an accurate and precise kinematic foot marker placement model. This will facilitate the study of 3D rearfoot and midfoot foot function during gait when using a Kinematic Motion Analysis System.

d. Providing invaluable information from images of lower limb alignment with barefoot, shoes and orthotic conditions taken in the upright position using a Positional MRI Scanner.

1.2. Thesis Objectives

a. To establish if a relationship exists between anterior knee pain and lower limb biomechanics.
b. To analyse the intratester reliability of a detailed clinical assessment.
c. To develop and test a foot marker placement model for use with the Vicon 370 Kinematic Motion Analysis System to analyse the biomechanics of the lower limb.
d. To develop and establish values for lower limb biomechanics in a group of asymptomatic subjects using a detailed clinical assessment and the Vicon 370 Kinematic Motion Analysis System.
e. To develop and establish values for lower limb biomechanics in a group of symptomatic subjects with anterior knee pain using a detailed clinical assessment and the Vicon 370 Kinematic Motion Analysis System.
f. To analyse the effect of lower limb alignment in shod subjects with anterior knee pain using the Vicon 370 Kinematic Motion Analysis System.
g. To analyse the effect of foot orthoses on lower limb biomechanics in subjects with anterior knee pain using the Vicon 370 Kinematic Motion Analysis System.
h. To investigate lower limb alignment on asymptomatic subjects using barefoot, shod and orthotic conditions on the upright Positional MRI Scanner.
i. To investigate lower limb alignment on symptomatic subjects with anterior knee pain using barefoot, shod and orthotic conditions on the upright Positional MRI Scanner.

1.3. Structure of the Thesis

Chapter 2 provides the reader with a comprehensive background and literature review on anterior knee pain and treatment particularly with the use of foot orthoses. Chapter 2 also reviews the measurement of the clinical tests performed in the clinical assessment and also the available literature on the Upright Positional MRI Scanner. Vicon 370 Kinematic Motion Analysis System is also reviewed and the marker placement models associated with it. Chapter 3 investigates the intrarater reliability of the measurements used in the clinical assessment by means of a pilot study. Chapter 4 introduces the foot model used in the motion analysis study and marker placement procedure. A pilot study is also included in this chapter which assesses the intrarater reliability of marker placement. In chapters 5 and 6 there are details of the randomised-controlled trial methodology for evaluation of the lower limb biomechanics, chapter 5 including the control group of asymptomatic subjects and chapter 6 including the group with anterior knee pain. Each trial is described in detail with specific reference to interventions, primary clinical outcome measures, clinical variables and the methods of statistical analyses. The results from each clinical trial are presented and discussed in detail. Chapter 7 investigates the effects of shoes and orthoses on lower limb alignment using the Upright Positional MRI Scanner. As with previous chapters, methodology, statistical analysis and a full detailed discussion are presented. Chapter 8 provides the reader with a discussion collated from conclusions from chapters 5, 6 and 7 and chapter 9 presents a conclusion and the clinical implications as a result of this research.

Chapter 2 – Literature Review

2.1. Introduction

This chapter reviews the past and current literature relevant to this field of work. A description of anterior knee pain, aetiologies and treatments available alongside a detailed review of foot orthoses with special emphasis on how they work and why they work is provided. The well documented coupling effect is examined in detail as this is particularly important in establishing if there is a direct relationship between anterior knee pain and lower limb biomechanics. A thorough review of the clinical measurements which are included in the clinical assessment is also provided in this chapter and an overview of the reliability of these measurements is given. The Upright Positional MRI Scanner and the Vicon 370 Kinematic Motion Analysis System are discussed as well as marker placement for the kinematic analyses.

2.2. What is Anterior Knee Pain?

There is no clear consensus in the orthopaedic literature on the terminology, aetiology and treatment of the anterior part of the knee (Cutbill et al 1997; Holmes 1998 and Thomee et al 1999). The term "anterior knee pain" (AKP) is suggested to encompass all pain related problems of the anterior part of the knee (Thomee et al 1995). The different terminology used and still widely discussed such as chondromalacia patella, patellofemoral pain syndrome, patellar pain syndrome, patellar pain, malalignment syndrome and patellofemoral arthalgia illustrate the complexity of AKP. These names are all often used synonymously with AKP.

Aleman (1928) when describing pathological changes of the retropatellar cartilage defined the term chondromalacia patella (CP) at the beginning of the 20th Century. Chondromalacia specifically means abnormal softening of the articular cartilage on the under surface of the patella and is widely used to designate the clinical syndrome of patellofemoral pain (Callaghan & Baltzopoulos 1992). McNichol (1986) questioned the nomenclature by stating that chondromalacia patella suggests a patellar and femoral softening of the articular cartilage, yet this pathological change is very rare in young athletes with patellar pain. Garrick (1989) stated that the term chondromalacia patella is inappropriate and pathologically incorrect and Witvrouw et al (2000) also agrees and reports that the term is a wastebasket term and should not be used synonymously with AKP. Fulkerson & Hungerford (1990) state that the term is now restricted to those instances where articular cartilaginous degeneration has shown to be present. Bourne et al (1988) suggested that the term should be discouraged and should only describe changes in the knee found on arthroscopy or athrotomy. They also stated that there are many causes of AKP such as biomechanical or biochemical and one cause could be a direct cause of another or biomechanical changes could cause symptoms on the articular surface similar to CP. However, there seems to be general agreement that there is a definite distinction between AKP and chondromalacia patella (McNichol 1986 and Deveraux & Lachmann 1984) but there seems to be a contentious issue surrounding the clinical features between AKP and CP (Wilson 1989).

According to La Brier (1993) and Baquie & Bruckner (1997), AKP is the most common complaint of runners seen in sports medicine facilities and Taunton et al (2002) state that the most common overuse running injury 20 years ago was patellofemoral pain; this is still the case to date.

It affects as many as one in four individuals in the general or sporting population (McConnell 1986; Anderson & Herrington 2003; Witvrouw et al 2003) and it can occur at all ages and in both genders. In the general population, it has been observed to be more prevalent in active females according to Sathe et al (2002), Kannus (1987), Whiteside (1980) and Gray et al (1985). However, Bennell & Crossley (1996) found that there was no significant difference between males and females and AKP. Zuluaga et al (1995) stated that in the athletic population the numbers are usually equal. One or both knees can be affected and it is characterised by retropatellar pain, felt specifically when the knee functions under load in flexion, classically when ascending and descending stairs, pain during and after physical activities mainly running, cycling or even squatting (Insall et al

1976). Running, whether competitive or recreational is the most common form of exercise undertaken and the majority of injuries incurred affect the lower limb (Neely 1998). Clement et al (1981) conducted a retrospective study of patients seen by one physician over a period of two years and reported that a total of 1650 patients were seen for 1819 running -related injuries. Korkia et al (1994) proved that 41% of all injuries amongst 144 British triathletes were overuse injuries and 65% occurred during running. Prolonged sitting with the knee flexed, often known as "movie-goer's sign", also causes pain because of the extra pressure between the patella and the femur but can be relieved upon extension of the joint (Juhn 1999). Steps, hills and uneven surfaces tend to aggravate AKP. Once started, AKP frequently becomes a chronic problem forcing the patient to stop sports and other similar activities (Witvrouw et al 2000). However, the problem is not always resolved by simply discontinuing these sporting activities as the pain that results often interferes with simple normal everyday activities such as walking or even standing (Craik & Oatis 1995).

2.2.1. Who are at risk- Men or Women?

Despite historical struggles, women today are participating in sport and exercise (Arendt 1996) although there are numerous unanswered questions regarding gender differences and AKP (Alexander 1998).

Bell et al (2000) conducted a study of Army trainees and found that women experienced twice as many injuries as men. However, this was thought to be because the women entered the training course at a significantly lower level of fitness than the men. They concluded that although the women were less physically fit than their male counterparts, they made much greater improvements in fitness over the given time.

Wider pelvis, menstrual irregularities, weaker musculoskeletal system, higher body fat and increased ligamentous laxity are all possible reasons as to why women could be more at risk of injury than men. Bennell & Crossley (1996) found that women with a history of menstrual disturbance had an increased risk of musculoskeletal injuries but did not mention the knee specifically. Lloyd et al

(1986) found similar results where female runners all of whom had injuries were more likely to have absent or irregular menses. Cavanaugh (1990) states that women are more prone to injury due to a higher body fat content than men and state that extra body fat can hinder performance and may contribute to injuries during the weight bearing phase of running.

The Quadriceps angle (Q-angle) is a frontal plane measurement formed by the intersection of a line from the anterior superior iliac spine (ASIS) to the centre of the patella and a line from the centre of the patella to the tibial tubercle (Moeller & Lamb 1997). Kantaras et al (2001) maintain that the normal value is 10 to 15 degrees for men and 15 to 20 degrees for women. This could mean that women have an immediate predisposition to AKP because an increased Q-angle may increase the laterally directed force on the extensor mechanism, predisposing the patella to mal-positioning and instability (Smith et al 2008).

2.2.2. What are the causes of AKP?

The exact aetiology of the condition has not been determined and it is quite apparent that more than one cause is responsible for this condition.

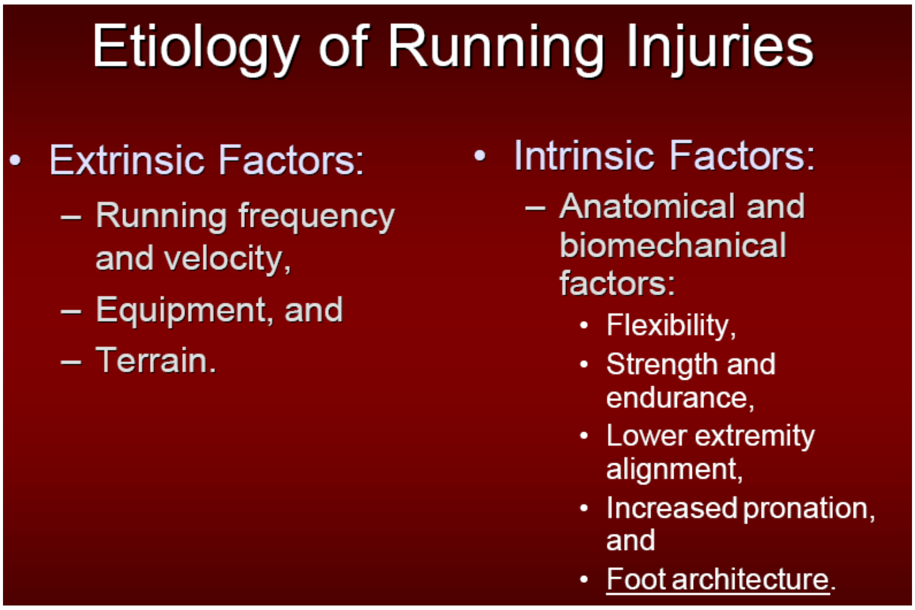


Figure 2.1 Aetiology of running injuries[1]

There is no consensus about the aetiology of AKP in the orthopaedic literature (Holmes 1998). However, by identifying factors which place an athlete at greater risk for injury, it may be possible to prevent future episodes and hence reduce injury incidence and prevalence. Risk factors are those entities which contribute to the occurrence of athletic injuries (Meeuwisse 1991). The risk factors responsible for AKP are traditionally divided into two main categories - intrinsic and extrinsic risk factors. The extrinsic factors relate to variables independent of the individual and environmental such as exercise load, type of exercise and the terrain on which the exercise is being conducted. The intrinsic category relate to the individual biological and psychosocial characteristics (Taimela et al 1990), such as gender, age and anatomical biomechanical factors.

There is continuing controversy between authors whether the intrinsic factors actually play a role in the aetiology of lower limb injuries. Nakhaee et al (2008) investigated if there was any relationship between the medial longitudinal arch height and knee injuries however they didn't specify which specific knee injuries. An independent *t*- test showed that the mean navicular drop value in healthy subjects was 5.3mm, which was significantly lower than that of the injured subjects who had a drop value of 7.4mm ($p=0.002$). However, despite the

[1] Hintermann and Nigg (1998), Williams et al (2001)

meaningful difference between the averages of navicular drop values, no relationship was found between the rate of knee injuries and the height of the medial longitudinal arch (MLA). Michelson et al (2002) also state that pes planus is not a risk factor for lower limb injuries in runners. Nakhaee et al (2008) concluded that extrinsic factors were more likely to determine the probability of an injury in athletes.

AKP can be divided by two mechanisms which are acute traumatic and overuse injuries. In case of injuries which are due mainly to sport, there is a higher incidence of acute injuries in contact sports, such as football, whereas in non-contact sports, such as running, there is a higher incidence of overuse related injuries (Armsey & Hosey 2004). Due to the enormity of both this review will only concern itself with the category related to overuse.

AKP can commonly be referred to as an overuse injury (Brukner & Khan 1993; Milgrom et al 1996; Finestone et al 1993; Thomee et al 1995 and Tria et al 1992). However, a more appropriate term may be "overload" because AKP can also affect inactive patients (Juhn 1999). Overuse injuries may be the most common class of sports injuries encountered by physicians (Francis et al 1997). They have been described in both civilian and military populations (Clement et al 1981; James et al 1978 and Taunton et al 1988). Baquie & Bruckner (1997) reported that overuse injuries at their centre during a one-year period were twice as frequent as acute injuries with the most common presentation being AKP. Clement et al (1981) also agreed that AKP is the most frequently encountered overuse injury. Running is one of the most common activities during which overuse injuries of the lower extremity may occur (Hreljac et al 2000 and Van Mechelen 1992). Various studies have reported that they are located at the knee and below for about 80% of all cases (Van Mechelen 1992 and Marti et al 1988).

2.2.3. What is an overuse injury?

Overuse injuries are clinically used to describe exertional pain and or dysfunction when there is no evident acute trauma involved (Rolf 1995). They occur when repetitive sub maximal stresses to bone or musculotendinous structures damages tissue at a greater rate than at which the body can repair itself (Krivickas 1997). The tissue is continually injured on the microscopic level and cannot repair itself as rapidly as the damage is being done. Muscle, tendon, ligament and bone are all able to undergo structural change and become conditioned to accept more biomechanical stress without suffering injury (Watson 1988). However, if these structures are subject to increasing repetitive sub- maximal stresses, the body cannot keep pace and injury may result. For example, during running the punishing stresses absorbed by the knee will render the area susceptible to injury if there are problems with muscle function or even minor biomechanical disturbances. These may be of no significance when the individual is involved in sedentary or non-repetitive activities (Taunton et al 1987). In most cases, this then produces degenerative changes leading to weakness, loss of flexibility and chronic pain (Nirschl 1992). Although it is common for athletes to suffer from overuse injuries, sedentary individuals have a substantial risk of developing an overuse injury. If this individual is starting out on a physical training program then "too much too soon" is a common problem and is responsible for up to 60% of all injuries (James et al 1978). Various epidemiological studies of recreational and competitive runners (Lysholm & Wiklander 1987, Macera et al 1989 and Jacobs & Berson 1986) have estimated that between 27% and 70% of runners sustain overuse injuries during any one year period. Clement et al (1981), Macera et al (1989) and Marti et al (1988) found that new inexperienced athletes are at an increased risk of injury than experienced athletes. The lower injury rate among more experienced individuals could be due to a musculoskeletal adaptive process that would decrease injury rate as the years of training increase. The level of force at which a given tissue can withstand may be increased gradually by increasing the training load, improving flexibility and strength, improving biomechanics and where possible, correcting anatomical malalignment (Krivickas 1997).

2.2.4. What is the treatment for AKP patients?

The amount of literature published on AKP and its management is a reflective measure of the frequency of its occurrence. Treatments vary enormously and a wide range of conservative interventions including rest, patellar bracing, ice, non-steroidal anti-inflammatory medication, electrical modalities, quadriceps exercises, foot orthoses and passive mobilisation of the patellofemoral joint, have been used singly, or in combination (Eburne & Bannister 1996). Conservative treatment historically has aimed to improve quadriceps muscle power to stabilise the knee and thus reduce pain. Whereas overall results achieved have been variable (Insall et al 1993). McConnell (1986) proposed a treatment programme designed to correct lateral subluxation and recorded success of 96% of cases. In this study it was concluded that the McConnell regimen did demonstrate a slight advantage to the isometric quadriceps exercises. Although it appears that it is a genuine advance, it is not the complete magic potion for anterior knee pain.

Foot orthoses and athletic footwear are commonly used rehabilitative interventions in addressing AKP. James et al (1978) and McKenzie et al (1985) noted that appropriate footwear and selective use of foot orthoses may be beneficial in treatment of individuals with AKP. Also, several authors have reported that a shoe alone may produce significant changes in lower extremity alignment (Cornwall & McPoil 1995; Bates et al 1979; Brown et al 1995). Cornwall & McPoil (1995) reported that shoes decrease maximum tibial internal rotation in subjects with pronatory-type feet. Bates et al (1979) found significant differences in rearfoot motion between shod and barefoot conditions. Brown et al (1995) found no significant differences in analysis of rearfoot motion between shod only and shod with an orthotic device.

Despite the high incidence and the extensive literature written, the exact cause of this disorder remains enigmatic although it is quite clear that there is more than one cause responsible for this condition. Much of the literature concerning aetiologies and risk factors is based on conjecture or expert opinion. This is especially true for lower extremity alignment, an area with a paucity of existing studies (Wen et al 1998). Lower extremity alignment is often discussed along with foot orthoses due to fact that biomechanical malalignment can often be corrected with them (Donatelli et al 1998). The next section discusses foot orthoses in detail and their implications to lower limb alignment.

2.3. Foot orthoses

Despite the fact that foot orthoses remain widely used, to the extent they are even used prophylactically in forms with similar functional purposes (motion control and stability running shoes), their mechanical function, consistency and permanence of effects are not specifically known. It has been suggested that positive outcome results may be due to mechanical and or proprioceptive mechanisms (Nurse and Nigg 2001). Recent biomechanical literature suggests that orthoses can produce kinematic, kinetic and muscle changes on healthy subjects (Mundermann et al 2003; Nester et al 2003). Human adaptation may play some role in the variance of results, yet no one has conducted a quantitative experiment applicable to a general pronated patient population.

2.3.1. What are foot orthoses?

Foot orthoses (FO's) are devices which are inserted between the foot and the shoe and have been used in clinical situations to modify lower limb biomechanics during the stance phase of gait during walking and running. Although the primary focus of orthoses research has been mostly at the rearfoot, they are frequently prescribed to relieve a wide variety of lower extremity ailments including knee pain (Mundermann et al 2001; Landorf & Keenan 2000; Clement et al 1981; Eggold 1981; Way 1999 and Eng & Pierrynowski 1993). Cutbill et al (1997) states that a common conservative form of treatment of anterior knee pain (AKP) is using foot orthoses. Eng & Pierrynowski (1993) also found that orthoses combined with an exercise programme can be an effective method in reducing the symptoms of AKP in young women. They also reported in a study in 1994 that medially wedged orthoses resulted in a decrease in the range of frontal and transverse plane motion at the knee during the contact and mid-stance phases of gait although the amounts were small and Nawoczenski et al (1995) also reported a decrease in the transverse plane motion.

They can be categorised into rigid, semi rigid and soft orthotics and are typically classified as non-posted or posted and non-moulded or moulded (Root 1994).

2.3.2. Who wears foot orthoses?

Foot orthoses are often prescribed to patients with excessively high or low arches to help correct abnormal weight bearing and gait conditions (Zifchock & Davis 2008). People with excessively high arches tend to have a diminished capacity for shock absorption due to increased stiffness and a smaller area for weight distribution (Zifchock et al 2006). Conversely, people with lower arches have a tendency to collapse into excessive pronation (Mann et al 1981). Both of these conditions can predispose the patient to injury. Kaufman et al (1999) reported that those with either high or low arched feet are nearly twice as likely to sustain a stress fracture as compared to those with normal arch height. Williams et al (2001) found that runners with high arches have an increased propensity for bony injuries while those with low arched feet have a higher rate of soft tissue injuries (Zifchock & Davis 2008).

2.3.3. Do foot orthoses work?

There has been a significant volume of literature published relating to them, some of which support their use and some which is either inconclusive or simply discourages their use. Although the mechanisms by which orthoses are sometimes effective are not fully understood, a significant reduction in lower extremity symptoms has been reported (Donatelli et al 1988; Sperryn & Reestan 1983). Current opinion is that the biomechanical effects produced by orthotics are not clearly understood and it has been suggested that any positive outcomes from using orthotics may be due to mechanical and or proprioceptive mechanics (Stacoff et al 2007).

As mentioned previously, the source of excessive motion of the rearfoot has been an extensive area of study (Razeghi & Batt 2000). It has been reported that medially wedged orthoses reduce the range of rearfoot pronation during the stance phase (McCulluch et al 1993) and the maximum angle of calcaneal eversion during the stance phase (Novick et al 1992). Stacoff et al (2000) found that wedged orthotics also reduce the range of internal tibial rotation associated with rearfoot pronation during running in symptomatic and asymptomatic subjects.

Mundermann et al (2004) found that different types of orthoses have different effects. The orthoses with the external postings would work better to decrease maximum foot eversion and the moulding type would work better to decrease maximum tibial rotation. They concluded that with a combination of moulding and posting the positive effects of the moulding would override the posting effects.

Researchers have gathered qualitative data from patient surveys to offer proof of foot orthotic efficacy (Gross et al 1991; Orteza et al 1992; Sperryn and Restan 1983). Surveyed patients either had completely or partially recovered from injuries such as shin splints, plantar fasciitis, stress fractures and inversion ankle sprains while using prescribed foot orthotics. In addition there have been several quantitative studies which demonstrate that foot orthoses affect both the kinetics and kinematics of gait when used by pronated subjects (Johanson et al 1994; Nawoczenski et al 1995; Novick and Kelley 1990). Payne & Davis (2004 unpublished work) examined the effects of three retail foot inserts on plantar fasciitis one of which was the same AOL orthotic device as used in this present study. The results showed all subjects except three had improvement in symptoms at one month (varied from 0% to 80%).

Unfortunately, the repeatability of quantitative results has been poor. Many researchers have been unable to confirm quantitative orthotic effects, or have found significant variations in effects. Foot orthotic therapy is therefore controversial, since quantitative researchers have been unable to repeatedly determine kinetic or kinematic effects. Very few researchers seem to understand that if there is an improvement in the subject's condition whilst using orthotics, how it actually happens.

Stackhouse et al (2004) found in their study (n=15) that rearfoot motion variables were reduced with orthotic intervention however the group average reduction was not large enough to be clinically significant and their sample was asymptomatic. More studies with larger numbers of subjects are needed as individual responses to orthoses appear to be highly variable.

Neptune et al (2000) conducted a study into the effects of an orthotic and vastus medialis oblique strengthening on knee pain. They concluded that the muscle strengthening yielded much more consistent results than the orthoses in reducing patellofemoral joint loading during running. They found that they had a beneficial effect in some subjects and no effect in others but never observed an adverse effect. Gross et al (1991) surveyed 500 long-distance runners who had been prescribed FO's. They reported that 76% of the subjects found complete resolution or a significant reduction of their symptoms and that 90% continued to use their devices even after their symptoms had resolved. James et al (1978) found that 78% of their treated runners were able to return to their original running programmes. Donatelli et al (1988) surveyed 81 subjects and the results showed 90% of them were completely satisfied with their devices.

Novick and Kelley (1990) found a substantial decrease in maximum and total pronation in subjects walking with shoes and orthoses compared with the same subjects walking with shoes alone. The authors however did not state the amount of posting or the location of the post used with each device.

Few studies have actually evaluated the use of orthoses over a sustained period of time. Stude and Brink (1997) reported an improvement in fatigue in a group of golfers over a 6 week period and Rome and Brown (2004) also reported significant improvement in a group of symptomatic and asymptomatic subjects (n=50) with rearfoot malalignment whilst wearing orthoses. They utilized prefabricated foot orthoses with 5-degree rearfoot varus posts in the experimental group and compared this group to the untreated group with the same foot alignment abnormality. They found no differences in postural control at baseline between the two groups. However, after wearing the foot orthoses for four weeks, the experimental group showed significant improvements of postural control in medial-lateral sway. The authors emphasized this was the first study on healthy patients (those with no ankle instability) with a specific foot deformity that showed the positive effects of orthotics to improve balance and postural control.

2.3.4. How do foot orthoses work?

To examine the biomechanical effects of the foot orthotic during gait, the effects of the orthotic on joints proximal to the ankle in addition to the motion of the foot and ankle should be studied. The mechanism responsible for their success is not very well understood (Ferber et al 2005). The main role of the orthotic, either functional or biomechanical, is to control excessive and potentially harmful subtalar and midtarsal joint movement during the stance phase of gait (Burns 1977).

They work in one of three ways according to Bowker (1987).

1. By preventing unwanted movement by stabilising or restricting the range of movement of one or more joints.
2. By correcting or controlling a deformity.
3. By totally or partially relieving body weight from a limb, joint or area of soft tissue.

Kirby (1992) stated that to be able to reduce or control the amount of pronation at the subtalar joint, an orthotic must increase the supination moment that is generated across the subtalar axis of the foot. The most obvious way to do this is to have the forces directed in an upward direction and located medial to the subtalar joint (STJ). Kirby (1992) also believed that since orthoses are in direct contact with the plantar surface of the foot, the most likely method in which a supination moment is produced is by generating a force at the medial plantar surface of the foot in an upward direction. Stackhouse et al (2004) report also that the mechanism by how FO's may reduce the symptoms of knee pain is not well understood. They found in their studies that mean knee flexion velocity decreased with FO intervention. This then results in reduced strain rates placed upon soft tissues at the knee joint thereby relieving knee pain.

Nester et al (2003) reported a considerable effect of an orthotic on the rearfoot complex. They noted a decrease in the mean value of 15 subjects walking with shoes only and walking with shoes and a medially posted rearfoot orthotic (31.2 degrees and 28.6 degrees respectively). This was however a 10 degree posted device.

Cornwall & McPoil (1995) and Nawoczenski et al (1995) have shown that in addition to reducing rearfoot pronation, they reduce internal tibial rotation. Stacoff et al (2000) also agree and found that medially wedged orthoses reduce the range of internal tibial rotation associated with rearfoot pronation during running. They did however use a symptomatic and asymptomatic sample. This is not a surprising discovery given the assumed coupling effect of the subtalar joint between the lower leg and the foot.

Previous motion-time studies have produced conflicting reports about the biomechanical effect of orthoses on the ankle and foot joints (Smith et al 1986). Although some of the discrepancies have arisen from the variation in the construction of the orthotic, the procedures used for the gait analysis may also have been a source of variability.

Although traditional research has focused on the use of orthotics to alter the gait cycle, recent literature has begun to concentrate more on the use of orthotics as an aid for proprioception and postural stability hence the lack of up-to-date literature on rearfoot and forefoot mechanics.

2.3.5. What is the coupling effect and how does it work?

Several researchers have documented the synergistic relationship between STJ pronation and prolonged internal tibial rotation during walking and running. Tiberio (1987) and Cornwall & McPoil (1995) reported a very strong (r=0.95) correlation between pronation and tibial rotation during walking and Nigg et al (1993) also observed a positive coupling (r=0.99) between them during running.

Pronation of the STJ is a tri-planar motion consisting of eversion, abduction and dorsiflexion of the calcaneus with respect to the talus (Donatelli 1993). It has been suggested that pronation occurs, in part, so that the foot can accommodate uneven surfaces better as well as attenuating shock (Isman & Inman 1969) and (Root et al 1966). During closed chain pronation, when the calcaneus is fixed to the ground it cannot abduct relative to the talus, therefore in order to obtain the transverse plane component of STJ pronation, the talus adducts or medially rotates. Due to the tight articulation of the ankle mortise, the tibia internally rotates as the talus adducts. During this cushioning phase of stance the knee joint flexes which is also associated with tibial internal rotation. Thus pronation, tibial internal rotation and knee flexion occur simultaneously (Buchbinder et al 1979 and Tiberio 1987). During the latter stance phase, the propulsive phase, the opposite actions occur and these motions reverse. The calcaneus inverts and the tibia and talus externally rotate as the knee extends. The rationale for studying joint motion and timing is based on the idea that asynchrony in these motions may result in injury (DeLeo et al 2004).

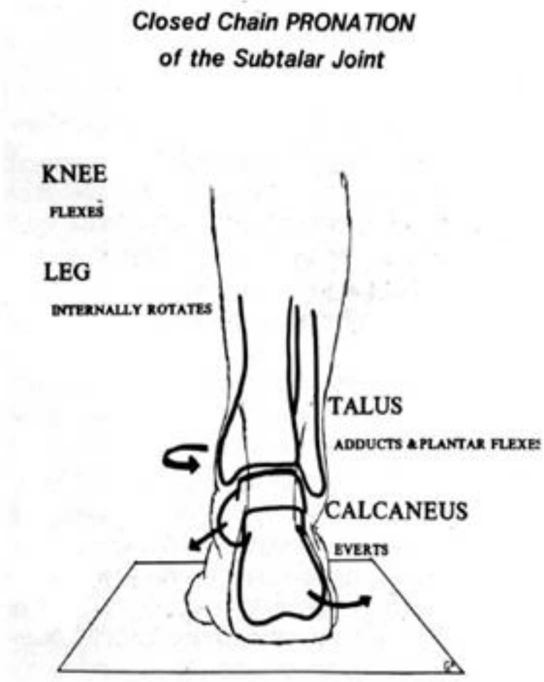


Figure 2.2 Closed chain pronation of the subtalar joint[2]

Tibial internal rotation is associated with knee flexion and rearfoot eversion and it has been postulated that these movements should be synchronous (McClay 2000). This means that peak knee flexion should occur approximately at the same time during midstance as peak rearfoot eversion and the reversal of these movements, knee extension and rearfoot inversion, should occur at approximately the same time after midstance (James et al 1978).

[2] McClay 2000

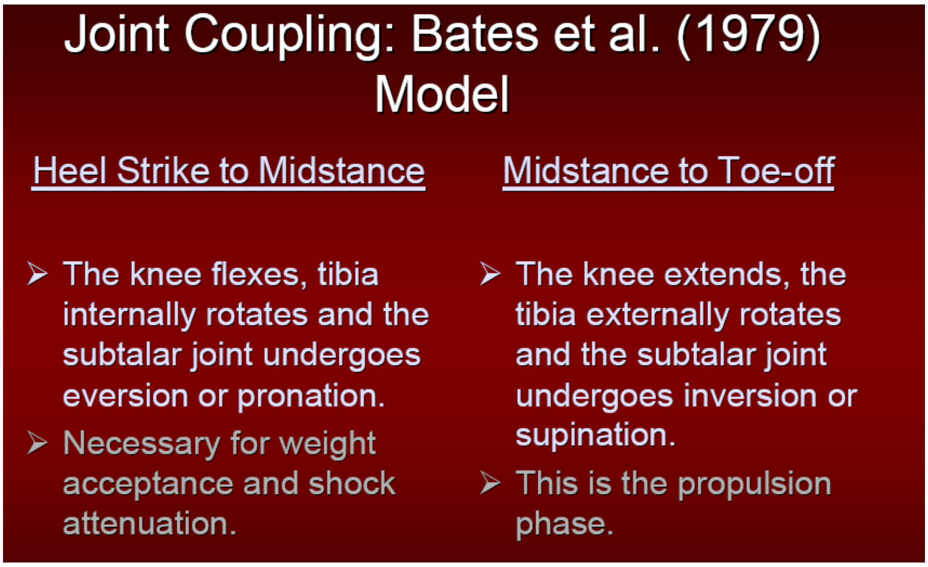


Figure 2.3 Joint coupling model[3]

The mechanism is such that if the STJ continues to pronate when the knee begins to extend, or if the STJ begins to supinate when the knee continues to flex, timing discrepancies between the joint actions would occur. It would appear therefore that the tibia would undergo antagonistic counter rotations at its proximal and distal ends which could lead to excessive stress at the knee joint and thus may result in injury. It has been suggested that abnormal rearfoot motion can lead to patellofemoral pain syndrome (Duffey at al 2000; Tiberio 1987).

2.3.6. How is coupling measured?

One way in which investigators have studied the coupling of eversion and internal tibial rotation is through the evaluation of the relative excursions. An eversion/internal tibial rotation excursion ratio (EV/TIR ratio) is formed by dividing the excursion of eversion by that of tibial internal rotation over the time period from heel strike to the respective peak values (occurring around midstance). As there is normally more eversion than internal tibial rotation, this ratio has been reported to vary between 1.0 and 1.8 (McClay & Manal 1997; Nawoczenski et al 1998; Stacoff et al 2000 and Willliams et al 2001).

[3] Bates et al. 1979

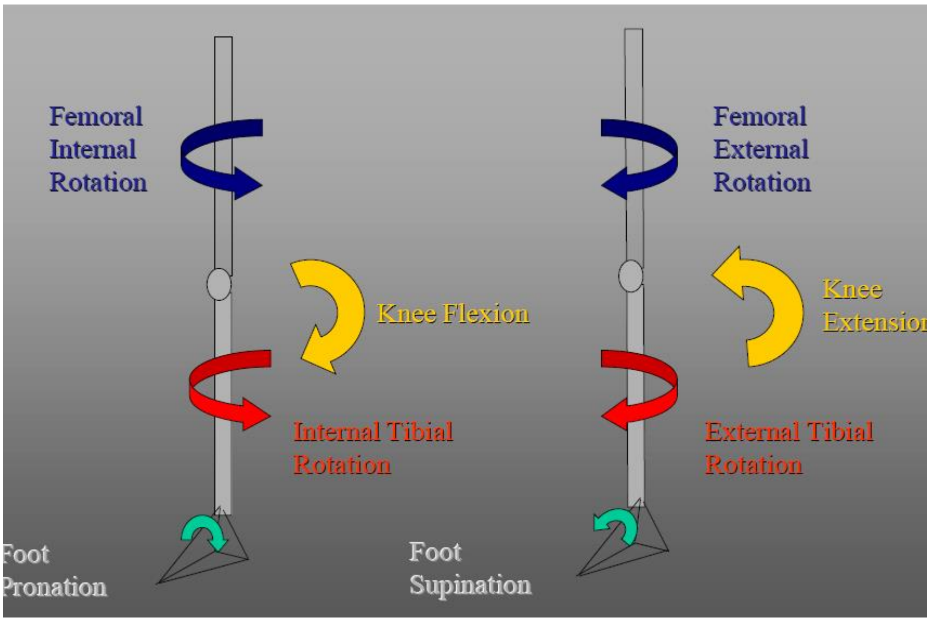


Figure 2.4 Diagrammatic display of joint coupling in the lower leg[4]

Tiberio (1987) suggested that the rationale for investigating the timing of joint motion is based on the idea that any asynchrony in the joint movements may result in injury. He proposed a mechanism for anterior knee pain that is related to abnormal joint coupling. He speculated that if pronation of the STJ is prolonged and continues beyond midstance, tibial internal rotation will also be prolonged. This then results in a mechanical dilemma at the knee, for knee extension begins around midstance and must be accompanied by tibial external rotation to maintain joint congruity. However, since the tibia is continuing to internally rotate with the talus, the femur must excessively internally rotate to obtain the relative knee external rotation needed. Tiberio (1987) also suggested that the compensatory femoral internal rotation may alter normal patellofemoral alignment and cause excessive contact pressures at the lateral facet of the patella.

The degree of coupling between the rearfoot and the knee is believed to be influenced by the orientation of the STJ axis in the sagittal plane (DeLeo et al 2004). However it is difficult to quantify the orientation of the axis without

[4] Tiberio 1987

invasive techniques. A number of authors have examined the relative amounts of both rearfoot eversion (EV) and internal tibial rotation (TIR) motion, which is suggestive of the orientation of the STJ.

DeLeo et al (2004) states that there is increasing evidence that arch structure influences EV/TIR ratios. As the arch height increases, the orientation of the STJ axis increases thus altering the ratio. A high arch may signify a higher internal tibial rotation and a lower EV/TIR ratio. Nawoczenski et al (1998), Stacoff et al (2000) and Williams et al (2001) showed that runners with low arches (thought to be related to excessive eversion) exhibited higher EV/TIR ratios.

Nigg et al (1993) reported that runners with high and lower arches exhibited similar rearfoot eversion excursion. However the runners exhibited greater internal tibial rotation excursions, resulting in lower EV/TIR ratios compared to the low arch group. Nawoczenski et al (1998) used radiographic measurements to classify arch structure and investigated differences in joint coupling in runners with high and low arches. Like Nigg and colleagues, they found no difference in rearfoot eversion between the two groups but found the high arch group exhibited greater internal tibial rotation, resulting in a lower EV/TIR ratio compared to the low arched group. McClay and Manal (1997) also compared EV/TIR ratios in runners who were excessive pronators and a group with normal rearfoot mechanics. They reported similar results as the previous authors. However, internal tibial rotation was greater in the pronator group resulting in a much lower EV/TIR ratio in this group (1.33) as compared to the normal group (1.42). Based on the majority of the evidence, it can be suggested that variations in the EV/TIR ratio can be attributed to internal tibial rotation excursion to a greater extent than rearfoot eversion excursion (DeLeo et al 2004).

Joint timing and joint timing differences are another simple way to measure joint coupling behaviour. This has been discussed at length in chapter 6 section 6.8.

2.3.7. Foot orthoses and the coupling effect

Many authors have speculated on factors which can increase a runner's risk of injury such as excessive pronation of the foot (Messier & Pittala 1988), increased internal rotation of the leg (Nigg et al 1993), increased knee adduction and external knee moments (Stefanyshyn et al 1999) and increased ankle inversion moments (McClay 2000). Therefore it has been suggested that by using an orthotic device, it can reduce the risk of injury by controlling these factors (Eng & Pierrynowski 1994; Nigg et al 1999; Nawoczenski et al 1995 and Mundermann et al 2001). However, due to earlier studies not finding any significant changes in any of the above factors, Hreliac et al (2000) suggests that reducing foot eversion and tibial rotation may not be the primary function of foot orthoses.

Foot orthoses are generally prescribed to control rearfoot eversion thus they will likely reduce the relative amount of eversion to internal tibial rotation and alter the joint coupling relationship (Ferber et al 2005). However, Ferber et al (2005) conducted a study to compare the joint coupling pattern and variability of the rearfoot and the tibia during running to have a better understanding of the mechanisms behind running injuries and the success of orthotics. They compared running in once injured but now treated (by an orthotic) subjects and non-injured subjects without orthotics. It was hypothesised that the treated runners would exhibit lower joint coupling angles without their orthotics compared to the controls. However, no significant differences were found. It was also hypothesised that there would be a decrease in the coupling angle between the orthotic and non-orthotic conditions and again, no significant changes were observed. The authors concluded by suggesting that foot orthotics do not produce significant changes in the rearfoot/tibial coupling.

Nawoczenski et al (1995) evaluated the effect of standard orthoses on the EV/TIR ratio of healthy runners with high and low arches. They reported negligible differences in eversion excursion for both groups when comparing the orthotic and non-orthotic conditions. They did, however, find a significant increase in TIR excursion in the orthotic group (5.2 degrees) as compared to the non-orthotic condition (8.1 degrees) in the high arch group. Therefore, orthotic devices can be responsible for significant changes in the EV/TIR ratios.

Although the exact aetiology of AKP is unknown, investigators propose that abnormal patellofemoral mechanics are the primary cause of it (Eng & Pierrynowski 1993). It has been discussed that although foot orthoses are frequently used in many clinical settings, there are many discrepancies regarding their usage and effectiveness.

It is anticipated that the results from the study (see chapter 6) will help answer and further open dialogue on such questions as, are the initially reported effects of foot orthotics, the true effects and should they be reported as such?

The following section discusses in detail the importance of accuracy and reliability when performing clinical measurements.

2.4. A review of the literature for the reliability of measurement

2.4.1. Introduction

Measurement of the limbs in a clinical examination is essential to determine baseline for treatment, to determine treatment efficacy and to assist in diagnosis. However, it has been the subject of much controversy and as yet there is no one single method that proves to be reliable or valid in its own right.

In order for a clinical measurement to have any scientific credibility, a reliable and valid clinical measurement technique must be employed. With regard to clinical assessment, reliability can be defined as the amount of agreement between successive measurements of the same joint by the same tester or different testers, namely, intratester and intertester reliability respectively. Validity is defined as the degree to which an instrument measures what it is supposed to measure; the extent to which it fills its purpose (Currier 1990). As such, it is clear that for a clinical measurement technique or tool to be useful, reliability and validity are fundamental pre-requisites.

Reliability and validity testing are usually performed to assess one of the following:

a) Instrumental reliability, i.e. the ability of the chosen instrument to perform reliably and will give reproducible results.

b) Rater reliability, i.e. the ability of the therapist administering the chosen instrument.

c) Response reliability, i.e. the reliability/stability of the variable being measured (Bruton et al 2000).

2.4.2. Goniometry

Goniometry is a technique commonly used in clinical practice for measuring the range of motion in body joints. The literature on goniometry is extensive since goniometers have been available for measuring joint range of motion (ROM) since the early 20th Century. However, examiners have not been able to agree on how to perform the correct procedures for goniometric measurements (Reese et al 2010). In 1965, the American Academy of Orthopaedic Surgeons published a manual of standardised methods of measuring and recording joint motion.

A variety of instruments are used to measure joint motion. These instruments range from simple paper tracings and tape measures to elaborate flexible electrogoniometers. A therapist may choose to use a particular instrument based on the instrument's accuracy, cost, availability and ease of use (Norkin & White 1985). The usefulness of goniometric measurements for providing objective assessments of a patient's initial status and progress depends on the reliability and validity of the measurements (Gogia et al 1987).

The universal goniometer (UG) is the most common instrument used when measuring joint motion. Moore (1948) designated this type of goniometer as "universal" because of its versatility. It can be used to measure joint position and ROM at almost all joints of the body. It is simple to use, non-invasive and inexpensive (Norkin & White 1988). However, despite its practical advantages, reliability is low due to inaccuracies that have been reported with improper alignment (Elveru et al 1988).

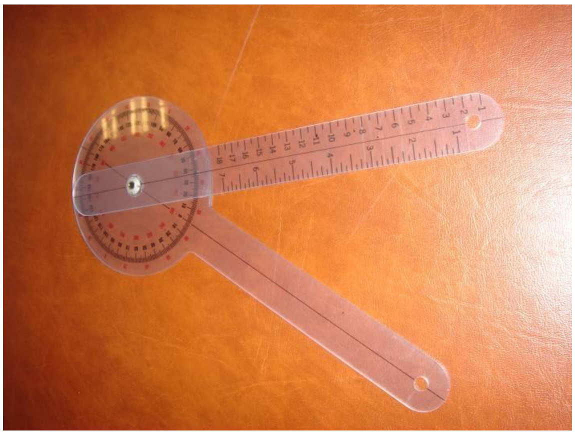


Figure 2.5 Universal goniometer

Electrogoniometers, introduced by Karpovich and Karpovich (1959) are used primarily in research to obtain dynamic joint measurements. They were thought to allow measurements to be made more accurately and to avoid erroneous joint recordings caused by misinterpretations of values obtained with the universal goniometer (Clapper & Wolf 1988). They are, however, expensive and take time to accurately calibrate and attach to the subject.

A less commonly used instrument is the fluid (bubble) goniometer, which has a fluid-filled circular chamber containing an air bubble and it works on a similar principle to a carpenter's level. This instrument may be easier to use and is therefore not subject to errors caused by alignment to bony landmarks such like the universal goniometer (Norkin & White 1995). However, disadvantages to the fluid-filled goniometer include an inability to measure outside the fluid's straight-plane movement in gravity (Rhealt et al 1988). There can also be variation occurring when measuring small joints or where there is soft tissue deformity (Miller 1985).

Another type of goniometer was designed and developed at The Rehabilitation Centre in Ottawa, as a result of the unpredictability of placing the fulcrum over the centre of rotation of the specified joint, called the Parallelogram goniometer (PG). The main advantage over the standard goniometer is its capacity to measure a joint angle without using a point representing the joint's centre of rotation (Brosseau et al 2001). Another benefit of the parallelogram goniometer is the shape as the range obtained is always represented by the angle between the two goniometric arms (Brosseau et al 2001). Brosseau et al (2001) performed an experiment, which examined the intratester reliability of the parallelogram goniometer and the universal goniometer when measuring active knee flexion and extension on healthy subjects with various knee restrictions. The results showed high intratester reliability with both goniometers using intraclass correlation coefficients (ICC's). The UG reliability was (ICC=0.977) in flexion and (ICC=0.985) in extension. Similarly, the PG reliability was excellent with (ICC=0.996 and 0.955) for flexion and extension respectively. They also looked at intertester reliability with both goniometers and concluded that although both were reliable, they recommended that the same therapist should take all measurements when assessing pathological knees.

Clapper & Wolf (1988) found excellent results when comparing a standard UG goniometer and an Orthoranger (a computerised digital goniometer). They found high ICC values ranging from the lowest result of (0.80) for hip adduction and the highest value of (0.96) for ankle plantarflexion.

A tape measure can also be used to measure ROM through observation of a change in distance from one segment to another. These observations are measured with a tape measure and are recorded in inches or centimetres.

2.4.3. Visual estimation

Visual estimation is used by some clinicians to assess joint range of motion in preference to using a standardised instrument especially when the subject has excessive soft tissue (Rowe 1964). The American Academy of Orthopaedic Surgeons (1965) suggested that where bony landmarks are not easily identifiable estimating the angle is as good as, if not better than, using a goniometer. However, there was no evidence offered to support this statement. Norkin & White (1985) however, do not recommend this procedure and Watkins et al (1991), Youdas et al (1991), and Low (1976) all report more accurate and reliable measurements when using a goniometer in comparison to visual estimates.

Brosseau al (2001) found that the data obtained from visual estimations revealed lower (r) values with the UG or the PG therefore the use of visual estimations is less valid than goniometric measurements. Croxford et al (1998) also is in agreement with the above-mentioned authors. They conducted a study comparing the intertester reliability of visual estimation and goniometric measurement of ankle dorsiflexion. It was demonstrated that intertester variation in visual estimation to be twice that of the universal goniometer thus concluding that a universal goniometer should be used to minimise measurement error between therapists when assessing ankle dorsiflexion in a clinical environment.

2.4.4. Reliability and validity

Establishment of reliability is essential for tests to be considered scientific and to be used with confidence (Rothstein 1985). Reliability and validity of measurement is a fundamental part of clinical practice, particularly when clinical assessment is based on subjective judgements for diagnosis, choice of potential intervention, and a review of management (Keenan & Bach 1996). For goniometry to provide meaningful information, an adequate measure must therefore be designed and tested for reliability and validity. This measure is tested for reliability. This is to ensure that repeat measurements on separate occasions (test-retest) and at the same time by the same therapist (intratester) or separate therapists (intertester) are reproducible. A goniometric measurement is highly reliable if successive measurements of a joint angle or ROM, on the same subject and under the same conditions, yield the same results. A highly reliable measurement contains little measurement error and a measurement that has poor reliability is not dependable and should not be used in the clinical decision-making process.

The validity of goniometric measurements has not been as extensively studied as reliability (Gogia et al 1987). Currier (1990) stated that validity is "the degree to which an instrument measures what it is purported to measure; the extent to which it fulfils its purpose". Most support for the validity of goniometry is in the form of content validity. Content validity is determined by judging whether or not an instrument adequately measures and represents the content-the substance-of the variable of interest (Rothstein 1985).

Some studies have examined criterion-related validity for various types of goniometer used in the clinical setting. Criterion-related validity justifies the validity of the measuring instrument in comparison to a well established "gold-standard" of measurement- the criterion (Norkin & White 1985). On a very basic level, an examiner may question the construction of a particular goniometer and consider whether the degree units of the goniometer accurately represent the degree units of a circle.

The best gold standard used to establish criterion-related validity of goniometric measurements is radiography. Gogia et al (1987) conducted a study to assess the intertester reliability of goniometric measurements at the knee and the validity of the clinical measurements by comparing them to measurements taken from radiographs. They measured the knee position of 30 subjects with radiography and with a large 360-degree plastic universal goniometer. Knee range of motion ranged from 0 to 120 degrees. Pearson product-moment correlation coefficients (r's) and intraclass correlation coefficients (ICCs) were used to analyze the data. The data analysis revealed high correlation and agreement - intertester reliability (r = 0.98; ICC = 0.99) and validity (r = 0.97-0.98; ICC = 0.98-0.99). Therefore goniometric measurement of knee joint position was considered valid. Enwemeka (1986) also conducted a study comparing goniometric measurements with radiographs and indicated a high degree of relationship between measurements obtained with radiography and goniometry. A study conducted by Brosseau et al (2001) sought to examine the criterion validity of the universal goniometer and the parallelogram goniometer and the criterion validity of visual estimations for active knee flexion and extension on subjects with various knee restrictions. They did this by comparing the results obtained with the UG and the PG with radiographs. The UG intratester reliability for flexion (ICCs = 0.997) and extension (ICC= 0.97 to 0.98). The results were also high with the PG (ICC = 0.99, 0.95- 0.95) for flexion and extension, respectively. The intertester reliability was high for flexion (ICC = 0.977- 0.982) and for extension (ICC = 0.893- 0.926) when using the UG. For the

PG, results ranged from (ICC= 0.959 to 0.970) for flexion and from (ICC= 0.856 to 0.898 for extension. Criterion validity varied from (r= 0.975 to 0.987) for flexion and from (r=0.390 to 0.442) for extension with the UG, and from (r=0.976 to 0.985) for flexion and (r=0.423 to 0.514) for extension with the PG. They concluded that intratester and intertester reliability were high for both goniometers. The results for the criterion validity varied. The study also revealed that it is preferable to use goniometry rather than visual estimations when measuring active range of motion. It is recommended that the same therapist take all the measurements when assessing range of motion for UG and PG goniometric measurements on patients with knee restrictions.

Despite the acknowledged need for accurate and reliable measurements, only a limited number of studies have been undertaken to determine the reliability of either the tester or the instrument (Moore 1949 and Low 1979) and the results varied considerably between these studies. Gogia et al (1987), Low (1976) and Ekstruand et al (1982) all studied goniometric joint measurements and found that joint range of motion can be measured with good-to-excellent reliability. However, several authors state that intratester reliability appears to be higher than intertester reliability (Boone et al 1978; Grohmann 1983; Low 1976 and Rothstein et al 1983). Boone et al (1978) express that the same individual should perform the goniometric measurements on the one patient in order to achieve reliable results.

The measurement of joint position and ROM of the extremities with a universal goniometer has been generally found to have good-to-excellent reliability. Gogia et al (1987), Grohmann (1983) and Hamilton (1969) have found that studies measuring a fixed joint position have higher reliability values than studies measuring ROM. This is expected because more sources of variation or error are present in measuring ROM than in measuring a fixed joint position.

2.4.5. Accuracy and precision

When conducting an experiment, the researcher must take steps to ensure the statistical test is reliable and valid, minimise measurement errors and only use measures that are precise and accurate (Gavin 1996). In all measurements, there is error. Knowing the error of measurement is important when determining the meaning of a change in values and is important in determining measurement precision. Precision refers to a measurement that has nearly the exact value each time it is repeated. Minimising the measurement error will increase the validity of drawing inferences from a study by rendering that study free of random error (Gavin 1996). Accuracy is also an important factor in measurement error. The accuracy of a variable is the degree to which it actually represents what it is supposed to represent. It can be involved in determining how well a variable measurement compares with another well-established measurement (Gold standard) (Gavin 1996).

Accuracy and precision can be wrongly interpreted. The following example of this could be a universal goniometer used to measure knee flexion and the procedure is repeated twenty times. If the readings are nearly equal on all the measurements, then this could be considered precise but not necessarily accurate. However, if the same measurements were compared to a Gold standard measurement such as a radiograph or a CT scan and both the readings were nearly equal, then this could be considered accurate but not necessarily precise.

Clarkson & Gilewich (1989) agree that the therapist should ensure that sources of error are minimised or do not occur when measuring. They state that one of the main errors to be avoided when measuring range of motion is reading the wrong side of the goniometer. Stratford et al (1984) also agree with this and state that if the goniometer arm was placed halfway between 40 degrees and 50 degrees, the tester may read 55 degrees instead of 45 degrees. Rome et al (1996) state other sources of error such as difficulty in reading the goniometer due to the graduations being too small or reading the measurement at an odd angle. Stratford et al (1984) stated that some goniometers may have received excessive wear and tear thus resulting in the increments being difficult to read. A different example of measurement error according to Clarkson & Gilewich (1989) is individual expectations of what the reading should be and allowing this to influence the reading. For example, if the study is evaluating intratester reliability, the tester often expects the repeated measures to be similar to that of the first value. Thus, the values of the intratester reliability may be uncharacteristically high. Stratford (1982 unpublished) found examples of this in a study conducted by himself whereby the intratester variation for the knee joint varied from a pooled standard deviation of 2.12 degrees (in cases which the goniometer scale was visible to the tester whilst taking the measurement) to a pooled standard deviation of 2.72 degrees (in cases which the goniometer scale was hidden to the tester whilst taking the measurements). Another example is end-digit preference. Testers have a tendency to read values that end with a particular digit (Rome et al 1996). Low (1976) proved this when he tested the reliability of visual estimation and goniometric measurements of elbow flexion. He found that when estimating the angle, only three out of fifty testers did not round the measurement up to the nearest 5 degrees, i.e. 5 degrees, 10 degrees or 20 degrees and only 14 testers did not round up when using the goniometer. Low (1976) also found that errors were made from placing the goniometer the wrong way round!

Clarkson & Gilewich (1989) report that taking successive measurements at different times during the day account for measurement errors. Stratford et al (1984) states that patients or subjects can have ranges of motion that not only varies within each day but from day to day as well. For example, patients with rheumatoid arthritis who may suffer from morning stiffness may give lower readings first thing in the morning for range of motion when compared to later on in the day. The environment also plays an important role. Unsuitable lighting or noisy establishments must be controlled in order to reduce the risk of error (Stratford et al 1984).

Measurement procedure errors such as different starting positions and testing methods are also important. Moore (1984) and Norkin & White (1985) have the same opinion that the reliability of goniometric measurements is affected by the measurement procedure. Watkins et al (1991) and Ekstruand et al (1982) found that intertester reliability improved when all the therapists used consistent, well-defined testing procedures and methods. Croxford et al (1998) agree that there is a need to heighten the awareness of the importance of adhering to well established and recognised procedures when measuring joint ROM.

Many studies of joint measurement methods have found intratester reliability to be higher than intertester reliability (Grohmann 1983; Boone et al 1978; Hellebrandt et al 1985 and Low 1976). Only a few studies found intertester reliability to be higher than intratester reliability (Defibaugh 1964). In all of these studies, the time interval between repeated measurements by the same therapist was considerably greater than the time interval between measurements by different therapists. This confirms what Clarkson & Gilewich (1989) reported that taking successive measurements at the same time increases the reliability factor. Moore (1949) also agrees with Clarkson & Gilewich (1989) in that a skilled tester may obtain inaccurate results by using a poorly constructed or unreliable measurement protocol and poor intertester reliability may result if therapists' used varied starting positions.

It is well documented that reliable goniometric measurements are difficult to obtain at some joints (American Academy of Orthopaedic Surgeons 1965). Boone et al (1978) found that upper extremity joint measurements are more reliable than those of the lower extremity. Some studies such as Brosseau et al (1997), Rothstein et al (1983) and Watkins et al (1991) have shown measurements to be more reliable for larger angles than for smaller angles. Moore (1949) and Rothstein (1983) found that in some instances, the axis of joint motion is either shifting continually or difficult to localise which may be due to the complexity of the joints according to Norkin & White (1995). Gajdosik & Bohannon (1987) stated that this could be an additional source of error in measuring joint range of motion. Hellebrandt et al (1945) also agreed that measurement error could be due to "peculiarities in anatomic structure or functional use".

Several investigations have been conducted to examine the reliability of using the mean of several goniometric measurements as compared with using just one measurement (Norkin & White 1985). Low (1976) recommends that using the mean of several measurements increased the reliability of the measurement. However, Boone et al (1978) and Rothstein et al (1983) found no significant difference and reported that one measurement taken by a therapist is just as reliable as the mean of repeated measurements.

Hellebrandt et al (1949) suggest that large goniometers are better than small ones but this theory has not been tested but Norkin & White (1995) suggest using large goniometers to increase the chances of reliability.

To conclude, the only way to obtain a high level of accuracy and reliability is for a therapist to fully understand the many different sources of error and variability and to standardise as many items as possible to reduce the number of variables in any investigation (Rome et al 1996).

The following section is a review of the past and present literature for the specific measurements which were included in the clinical assessment.

2.5. Clinical assessment- A review of the literature for measurements included in the clinical assessment.

2.5.1. Quadriceps angle (Q-angle)

Abnormal patellar tracking has been documented as one of the possible causes of AKP (Powers et al 1995). Generally, subluxation involves excessive lateral displacement of the patella between 10 degrees to 20 degrees of flexion (Fulkerson & Hungerford 1990). The natural tendency of the patella to track laterally has been described by Fulkerson & Hungerford (1990) as "the law of valgus". This is a consequence of the valgus orientation of the lower extremity where the relationship of the anterior superior iliac spine (ASIS) of the pelvis to the midline of the ground forms a 10 degree angle between the femur and the tibia (Powers et al 1995). Since the quadriceps follows the longitudinal axis of the femur, the angle is formally known as the quadriceps angle (Q-angle).

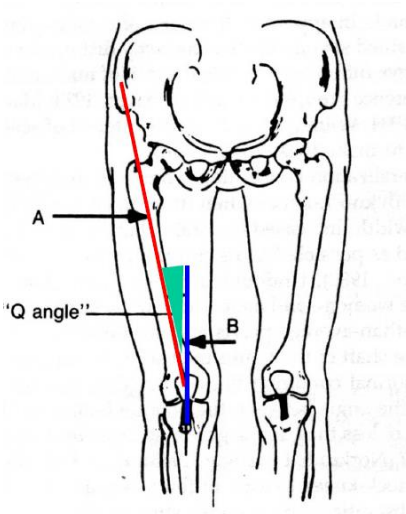


Figure 2.6 Quadriceps angle (Q-angle)[5]

The Q-angle is defined as the intersection of lines drawn from A- ASIS to mid patella and B-Tibial tubercle to mid patella as shown in Figure 2.6.

The Q-angle is an index of the vector for the combined pull of the extensor mechanisms and the patellar tendon (Tsujimoto et al 2000). It is a frontal plane measurement formed by the intersection of a line from the ASIS to the centre of the patella and a line from the centre of the patella to the tibial tubercle (Moeller & Lamb 1997).

There are many different methods used to measure the angle, although few studies using the Q-angle actually give detailed descriptions of their methods (Caylor et al 1993). The Q-angle can be measured in both the supine and weight-bearing positions as well as with the knee in both flexed and extended positions. Hossler & Maffei (1990) conducted two studies, one with the subjects' supine and one with the subjects' standing and found different values up to 5 degrees in each test. Caylor et al (1993) studied the reliability of the various methods of measuring Q-angle and found no significant difference in intratester Q-angle values between the extended and flexed knee positions ($p > 0.05$). With the knee in an extended position, intratester Q-angle ranged from (ICC= 0.84 to 0.90, and standard error of measurement (SEM) values ranged from (2.01 to 2.23 degrees).

[5] Cavanagh 1990

With the knee flexed, the intratester (ICC = 0.83) for both testers, and SEM values ranged from (0.68 to 2.45 degrees).

The Q-angle has been defined as a measurement to quantify lower extremity alignment (Hamill et al 1999). The rationale being that a higher Q-angle changes the contact and pressure patterns in the patellofemoral joint, leading to excessive pressure in locations that are not usually exposed to these stresses (Cox 1985). Smith et al (2008) testify that an increased Q-angle may increase the laterally directed force on the extensor mechanism, predisposing the patellar to mal-positioning and instability.

Skeletal malalignment has been shown to have a profound effect on the degree of the Q-angle. An increased Q-angle is often present when rotational malalignment of the femur and tibia are present. Such examples include femoral anteversion, genu valgum, tibial torsion and the position of the tibial tubercle (Insall 1976 and Paulos 1980). The tibial tubercle is affected by the tibial rotation, which in turn can be affected by the position of the foot. Patients may demonstrate one or several of these malalignments which may contribute to patellofemoral dysfunction. Tiberio (1987) demonstrated in a theoretical model that as the foot undergoes pronation, the tibia will internally rotate thus the tibial tubercle will be displaced in a more medial direction. As a result of these movements, the Q-angle should decrease. However, Corley et al (1997) states that the model from Tiberio (1987) does not take into account femoral or pelvic movements which occur during the walking cycle. Olerud and Berg (1984) disagree with Tiberio's (1987) claim and reported that they found a decrease in the Q-angle but when the tibia was externally rotated with the foot supinated. Olerud & Berg (1984) and D'Amico & Rubin (1986) both state that with pronation of the foot, internal rotation of the tibia occurs causing medial translation of the patella and an associated increase in the Q-angle.

Many authors hypothesise that an increased Q-angle plays an important role in AKP although there are differences of opinion (Chapman 1997). Shambaogh et al (1993) and Nadeau et al (1997) found in both their studies that those sustaining AKP had a larger Q-angle than those with uninjured limbs did. Duffey et al (2000) conducted a study examining the differences between injured runners with

AKP and non-injured runners. Although when tested, the Q-angle was not a significant discriminator between the groups however the injured subjects did have a slightly higher Q-angle value than the control group. There were a couple of methodological flaws that may have influenced the values. Although the tests were conducted on both legs, three times, nothing was written about the time interval in-between these tests therefore the tester may have been influenced by previous results. There was also no mention of how many testers conducted the experiments and the experience of the ones who did. Another limiting factor affecting the reliability of the Q-angle measurement could be that the authors never stated in their methods whether the Q-angle test was done weight-bearing, supine, knee extended or knee flexed. As mentioned previously, there can be differing results depending on how the angle is tested (Hossler & Maffei 1990). Moeller & Lamb (1997), Kannus (1992), Kernozek (1993) and D'amico (1986) all agree with this although Caylor et al (1993) and Thomee et al (1995) both found no difference in Q-angles between injured and non-injured subjects. Fulkerson & Hungerford (1990) consider there is no direct correlation between a high Q-angle and AKP however, Thomee et al (1995) state that a high Q-angle may be a contributing factor in maintaining AKP once it has been acquired but this needs further research.

In 2001, Sanfridsson et al compared clinically measured Q-angle values with radiographically measured Q-angles in patients and healthy knees. The results were striking in that the Q-angle was not increased in any of the patient groups but instead there were decreased Q-angles in the affected knees.

There is a lot of controversy over the size of a normal Q-angle and what is abnormal. The Q-angle is regarded as normal with the quadriceps relaxed if it is between 13 and 18 degrees (Reid 1992; Insall et al 1976 and Fulkerson 1983) but also state that these values differ between practitioners. Tomisch et al (1996) agree and state that the method of measuring the Q-angle differs between practitioners therefore differing values will occur. However, Geriniger (1999) thinks that the Q-angle is normal if it is between 5 and 15 degrees.

Schuster (1978), Anglietti et al (1983), Alexander (1998) and Woodland & Francis (1992) all report that women typically have greater Q-angles than men. They all state that 17° is normal for women and 14° for men. However, La Brier & O'Neill (1993) and Hughston et al (1984) state that anything more than 10° in men and 15° in women is abnormal. The Manual of Orthopaedic Surgery (1972) considers angles greater than 15° degrees in men and 20° degrees in women pathological. Kantaras et al (2001) maintain that normal value is 10° to 15° degrees for men and 15° to 20° degrees for women.

The inability of researchers to reach a consensus on what might be considered normal has raised doubt about its diagnostic value (Livingstone & Mandigo 1999). The lack of a standardised measurement protocol is at least partially to blame since the size of the angle can differ dramatically when measured using different procedures (Woodall & Welsh 1990). Indeed, variations in subject posture (Woodland & Francis 1992), knee position and foot position (Ando et al 1993) can all alter the size of the angle therefore direct comparisons between studies using differing measurement procedures are therefore not feasible (Caylor et al 1993).

2.5.2. Ankle dorsiflexion and plantarflexion

An adequate range of motion at the foot and ankle is a necessary component for the performance of functional activities such as running, ascending and descending stairs and many more weight bearing sporting activities which require the flexed knee to move forward over the foot when in contact with the ground (Bennell et al 1998; Donatelli 1996). Limited ROM in plantarflexion and dorsiflexion of the ankle has been suggested to be associated to running injuries (Hreljac et al 2000).

Ankle joint dorsiflexion is carried out routinely as part of a static lower extremity examination for the diagnosis of ankle equinus (Rome et al 1996). Proper ankle function, especially the dorsiflexion component, is essential for normal gait (Root et al 1977). When there is limitation of ankle dorsiflexion this is called ankle equinus. The effects of ankle equinus, with contracture of the Achilles tendon, may have a negative effect on foot and ankle function (Saxena & Di Giovanni 2011). They can involve the leg, knee, pelvis and spine and even the upper body. Acquired shortness of the posterior crural muscles, which produce ankle equinus, can result from many sources. Davis' Law states that "soft tissue under prolonged tension will elongate" and the "same is true in retrospect". Soft tissue will contract to its transformed position and thus this is the most frequent underlying aetiology of acquired muscular ankle equinus.

Downey (1992) cites that repeated use of high-heeled shoes, Achilles tendon (TA) rupture, or prolonged casting with the foot in a plantarflexed position can create an equinus position at the ankle joint. Limited ankle dorsiflexion (ankle equinus) has been associated with lower limb pathology (Hill 1995). Limited ankle dorsiflexion is thought to necessitate midtarsal joint dorsiflexion thus leading to midfoot collapse. Fortunately, the body will compensate for ankle equinus by flexing the knees, early heel-lift, foot pronation, and foot abduction. It may also include hyperextension of the knees or flexion at the hip (Hill 1995). The term "compensation" implies something that counterbalances or makes up in one place for something that is lacking in another (Oxford 2010).

Tiberio et al (1989) conducted a study to determine whether a significant difference exists between subtalar joint neutral position (STJN) and subtalar joint inversion position when measuring ankle dorsiflexion. They found that inversion elicited ten degrees greater dorsiflexion than when in STJN position.

The position of the subject may also influence the amount of motion available. The most widely used position is the supine or prone position but other authors have suggested either a seated or standing weight bearing position (Baggett & Young 1993). Baggett & Young (1993) evaluated ankle dorsiflexion in a static weight bearing position where the subject stood upright facing a wall and was asked to lean forwards. With the knee extended, a small wedge was placed under the head of the first metatarsal in order to prevent eversion of the rearfoot. However, Gross and Niapoli (1993) reported that this method was not appropriate and represented a false measurement, as the joint actually being measured is the midtarsal joint. Jones et al (2005) undertook a study examining the intraobserver and interobserver reliability of ankle dorsiflexion in a weight bearing position using a measurement tool which uses a linear distance method to establish the distance that the knee can move forward over the foot which is fixed to the ground. They found far more agreement and higher ICC results with the injured group that the asymptomatic group. They also found better reliability with independent observers than the same observer which is surprising. Bennell et al (1998) also conducted a similar study using the weight bearing lunge test to measure ankle dorsiflexion and found similar results. They however found excellent intrarater results as well as interrater results. They concluded that differences in skill level did not influence the repeatability of the results therefore may be an easier method of measuring ankle range of motion. Although this seems like an easy way to measure it both for the patient and to get good results, it must be remembered that it does not measure actual motion at one specific joint. Dorsiflexion, as measured by this test is a combination of movement at a number of joints including the talocrural, midtarsal and subtalar joint. Another negative point to this method is that there may be a number of patients who cannot perform this movement and weight bearing may be contraindicated.

Several authors have reported the "normal" ranges of motion for the ankle joint and they vary greatly (Elveru et al 1988). Normal ankle ROM is traditionally thought to be 10 degrees of dorsiflexion (Root et al 1979) but the American Academy of Orthopaedic Surgeons (1965) state that the normal range is anything from zero to 20 degrees as does the American Medical Association (1990). Other studies report a range from 8 to 26 degrees (Rome 1996). This wide range of variability presents a dilemma for therapists who try to assess the normality of subject's range of motion (Oatis 1988). Normal ankle plantarflexion ROM is traditionally thought to be 50 degrees (American Academy of Orthopaedic Surgeons 1965) but the American Medical Association (1990) state that the normal is 40 degrees but these values are based on a method that uses visual estimation. Grimston et al (1993) suggest that ankle joint ROM is age-dependent and generally maximum ankle joint ROM occurs before 20 years of age with decrements beginning as early as 17-20 years of age. However, even though Rome et al (1996) agree with the fact that there is a general decrease in ankle dorsiflexion from early adulthood, they report an age range of 29-39 years.

The reliability of goniometry (as discussed previously in this chapter) is dependent upon standardised measurements (Elveru et al 1988 and Youdas et al 1993) and ankle joint ROM measurement is particularly difficult to standardise due to the complex motion at the joint (Engsberg 1987). The hand-held universal goniometer remains the most widely used instrument to measure ankle joint ROM (Rome et al 1996), however, there are many sources of measurement error which exist when measuring it (Wright & Feinstein 1992). The coincidental location of the ankle joint fulcrum with the axis of rotation, and the multi-axial movements of the ankle joint have been reported by Elveru et al (1988) to present sources of measurement error. Clinical measurement of ankle ROM may be moderately reliable if conducted by the same therapist over a short period of time (Youdas et al 1992). Elveru et al (1988) investigated intratester and intertester reliability of goniometric measurements at the ankle joint. They found good intratester reliability for both plantarflexion and dorsiflexion and adequate intertester reliability for plantarflexion. However, intertester reliability for dorsiflexion was poor. This suggests that perhaps measures for dorsiflexion should be re-evaluated or to identify if the presence or absence of plantarflexion is more reliable and would suffice for clinical examinations. However, the results of the study should be interpretated carefully as several methodological flaws were identified. Only subjects with orthopaedic or neurological conditions were included in the study therefore does not represent a normal sample of the population and no standardised protocol was made for measurements at the ankle joint. The tibia and the ankle joint were not held in stabilised positions, therefore additional motion may have influenced the reading thus influencing reliability. The study, mentioned above by Tiberio et al (1989), also has flaws as the study only included eighteen women, which is a very small sample size and the chance for sampling error is indirectly proportional to the size of the sample (Polgar &Thomas 1991). The women selected had no history of congenital or traumatic conditions. Boone et al (1978) also agree from their study that intratester reliability is much higher when using the goniometer for ankle joint ROM than intertester reliability.

Rome et al (1996) investigated the reliability of three types of goniometer on ankle ROM and found poor interdevice reliability both among and within therapists. This suggests that when goniometers are interchanged in a clinical setting, which is possible, particularly when the patients are seen by different therapists each visit, there is low reliability of measurements. They concluded by stating that error associated with using multiple devices is much higher than when using a single device. Hellebrandt et al (1949) strongly agree with this and found that reliability was improved when all therapists in one department used the same goniometer. Rome et al (1996) also agree that once a therapist has chosen a particular instrument that therapist should use that particular instrument, and not interchange with any other instrument or therapist, but, Rothstein et al (1983) found in their study that for all measurements of knee and elbow movements, the three goniometers used could be used interchangeably.

2.5.3. Limb length discrepancy (LLD)

Leg length discrepancy is a common problem found in as many as 40% (Subotnick 1981) and 70% (Woerman and Binder-Macleod 1984) of subjects where paired limbs are noticeably unequal (Gurney 2002). To date, there is no universally accepted clinical method for measuring leg length (Ingram 1980). The degree of limb length difference (LLD) that is clinically significant remains controversial (Beattie et al 1990) as does the acceptable amount of LLD necessary to warrant treatment. Subotnick (1981) reported that a difference of as little as 3mm is significant whereas Anderson has stated that a difference of less than 19mm is acceptable.

Limb length difference is widely criticised as to whether or not it plays a role in the aetiology of AKP (Kannus 1992). Gogia & Braatz (1986) stated that any problems resulting from LLD might depend on the degree of inequality. Some authors have claimed that a limb-length discrepancy leads to mechanical and functional changes in gait (Morrissey 1990). Treatment has been recommended for discrepancies of less than one to five cm (Gross 1978; Morrissey 1990).

Brody (1986) claims that an LLD of 6mm or greater can cause injuries to runners although, Wen & Puffer (1997) conducted a study that found leg length discrepancy was not an important risk factor in runners with knee pain however the methodology had limiting factors. The subjects were chosen from a cohort of low mileage runners, which is not indicative of a representative population. There was also a potential for selection bias known as "the healthy effect" (Marti et al 1988). Athletes who were already injured at the time of selection may not have wanted to volunteer for the study and athletes whom were prone to injury in the past may also not have wanted to volunteer for fear of re-injury. Another limiting factor was the lack of reliability of the measurements. Intratester reliability results were of limited use due to the fact that the testers performed the second test immediately after the first one and therefore were not blinded to the results of the first test. This may influence the testers' results especially if they were inexperienced which in this study they were. Intertester reliability was not checked due to the lack of time available.

There is lots of criticism on the reliability and accuracy of leg length measurement (Freiberg 1982; Mosseley 1987; McCaw et al 1991). Although associated with low reliability and validity, the tape measure remains the most common instrument used to evaluate LLD (Freiberg et al 1988). This is even though the reported validity and intratester and intertester reliability coefficients of tape-measured methods are consistently less than radiographic methods (Clarke 1972). Nichols & Bailey (1955) also agree that by taking measurements from radiographs to determine leg length and then calculating the difference is generally considered the most accurate method of testing LLD. They also state however, that because of their cost and the fact that they expose the subject to the adverse effects of radiation, radiographs are impractical and unjustifiable for determining LLD.

Although the tape measure method is easy to administer and costs very little (McCaw & Bates 1991), there are many potential sources of error with the method. Differences in the circumferences of both legs could contribute to distance differences, as could unilateral deviations along the long axis of the leg such as genu valgum or genu varum (Beattie et al 1990). McCaw & Bates (1991) think that tape-measurement inaccuracy stems from its reliance on bony prominences such as the ASIS and medial malleolus. Beattie et al (1990) also state that any asymmetries in the surface contours of the skin at the thigh, knee, or lower leg may alter the position of the tape measure therefore leading to inaccurate LLD measurements.

Beattie et al (1990) reported that measurements obtained with a tape measure appear to be valid for assessing subjects with LLD when the mean of two measurements is used. They conducted a study using two examiners who obtained repeated measurements of LLD on 50 subjects. The results showed good intratester reliability (ICC =0.807) and fair intertester reliability (ICC =0.668) when comparing the first measurements obtained by each examiner. When they compared the mean values of paired measurements, the intertester reliability rose significantly (ICC=0.910).

Ideally both limbs should be of equal length but a certain amount of discrepancy can be considered normal although authors differ on what the criterion for normal is and when a discrepancy should be treated.

A leg length inequality of less than 3cm is classified as mild and moderate differences include leg length inequality between 3 and 6cm. Differences greater than 6cm is classified as severe (McCaw 1992).

2.5.4. Subtalar Joint Neutral (STJN)

The definition of STJN has important implications for assessment of foot function. It is the most widely used reference point for the clinical measurement of the relationship of rearfoot to forefoot (Pierrynowski & Smith 1997). It provides consistency in positioning the foot before assessing structural or bony deformities of the foot and leg (Oatis 1988). It also provides the clinician with a relative zero point from which to measure STJ range of motion and serves as a point of reference for other lower limb measurements. The ability to identify the STJN, therefore, would appear to have important implications for assessment of the lower extremity (Levangie & Norkin 2001). However, there are differences in opinion in the literature as to the correct definition of STJN position and the most reliable and valid form of measurement. In 1964, Wright et al defined the subtalar neutral position as *"the position of the subtalar and talocrural joints when the subject was standing relaxed with the knees fully extended, the arms at the sides, feet six inches apart and with a comfortable amount of toeing-out."* However, McPoil & Cornwall (1994) argue that this definition describes the relaxed calcaneal stance position today and not the STJN position. The American Academy of Orthopaedic Surgeons (1965) state the STJN position is that position where the longitudinal midline of the leg and heel bisection are parallel although Elveru et al (1988b) does not agree and finds this definition not clinically useful because it does not take into account anatomical differences between subjects. Merriman & Tollafield (1995) define neutral position as the foot being neither pronated nor supinated. LaPointe et al (2001) maintain that any deformity such as tibial varum or rearfoot valgus makes this definition irrelevant. They give an example of a patient with a fully compensated rearfoot varus who may be at the end range of eversion when the calcaneus is aligned with the tibia therefore the definition cannot be applied universally.

Root et al, in 1971 defined subtalar neutral as the position of the STJ that was neither pronated nor supinated and in 1977 defined it, as the position in which the forefoot is locked on the rearfoot when the midtarsal joint is maximally pronated. He also classified it, as the point from which the calcaneus will invert twice as many degrees as it will evert. This position is generally determined by palpation of the talus on the calcaneus. These are only a few examples and Root's theories on biomechanics have been the gold standard for many years. However many of the current researchers such as McPoil & Cornwall (1994), Elveru et al (1998) and Pierrynowski & Smith (1996/1997) are discarding his ideas in favour of new theories about STJN position and the biomechanics of the foot during the gait cycle.

The position of the STJN is also important in casting and for orthoses fabrication (Elveru et al 1988b). It is the position typically used by therapists to obtain a cast representation of a patient's foot before fabrication of biomechanically functional orthoses therefore it is important for therapists to agree as to the true definition of STJN position. As mentioned above, McPoil & Cornwall (1994) state that the neutral position of the rearfoot during gait is the resting calcaneal stance position, not the subtalar joint position. They even suggest that due to this finding, the fabrication of orthoses in STJN position may be contraindicated however they do not explain why there is evidence to show that orthoses fabricated from a STJN position can improve patient symptoms. Donatelli et al (1988) found 91% improvement in patients' symptoms when supplied with orthoses casted in a STJN position.

Eng & Pierrynowski (1994) found that the correction of structural foot abnormalities with soft foot orthoses resulted in small changes in frontal and transverse motion at the knee. Although the sample size was small, female and all adolescents, these results show that there is a clinical relationship to orthoses and patellofemoral pain syndrome.

Astrom & Arvidson (1995) state that when carrying out biomechanical evaluations or prescribing orthoses, a gold standard must be set as a reference. This gold standard is the "ideal foot". It is based on the theory that it will be more efficient and provide optimal function with minimal risk of injury if granted a balanced range of subtalar joint motion. Astrom & Arvidson (1995) also state that a vertical stance position of the calf and the calcaneus in the STJN position is the 'ideal foot' and provides optimal function with minimal risk of injury. However, none of the subjects conformed to the "ideal foot" theory, which should be abandoned in favour of a reference based clinical observation, rather than theoretical observations. McPoil et al (1998) and Garbalosa et al (1994) also both conducted studies involving foot position of healthy subjects and McPoil et al (1998) found that out of 116 feet surveyed, not one was an "ideal foot" in STJN position. The latter study found that out of 234 feet sampled, 90% had a forefoot condition bilaterally. These studies agree with Astrom & Arvidson (1995) and McPoil et al (1999) stated " if so many feet are abnormal in a healthy adult symptom free population, then one could question whether the criteria used to define a "normal" foot structure is even appropriate." However, the term "normal" has been used in as many different contexts as there have been people whom have written about it (Phillips 2000). To every individual, the term has a different meaning. Root et al (1971) defined the normal foot as

"A set of circumstances whereby the foot will function in a manner which will not create adverse physical or emotional response in the individual. This applies when the lower extremity is used in an average manner and in an average environment, as dictated by the needs of society at the moment."

This position has been defined differently by various investigators, with some issues raised as to the appropriateness of the measurement techniques (Levangie & Norkin 2001). Since small angular deviations may be of clinical relevance, the accuracy and consistency of measuring the STJN position is crucial in the management of foot and leg disorders. Therapists should be proficient in performing this skill and should be accurate and consistent. The method described above by Elveru et al (1988) was used in a study by Astrom & Arvidson (1995) and they found the average position of the calcaneus in the palpated STJN position to be 2 degrees of calcaneal valgus. However, in another study by McPoil & Cornwall (1994) also conducted using the "Elveru" method found an average STJN position of 1.5 degrees of calcaneal varus. The reliability of this method varies according to experience of the clinician and position of the patient (Pierrynowski & Smith 1997).

A study by Ball & Johnson (1993) investigated the use of a flexible electrogoniometer on the reliability of STJN measurements. They showed a high intertester variation of 7 degrees for STJN whereas the intratester variation was only 2.5 degrees. However, this study was conducted with a small sample size and the outcome could have been influenced by instrumentation that was not blind to the tester (Rothstein 1985 and Clarkson & Gilewich 1989). This phenomenon is known as the Rosenthal Effect whereby the expectations of the tester influence the outcome of the study (Polgar & Thomas 1991). It is also not clear whether the design of the study was randomised which if not, would further invalidate the conclusions since a randomised sample is more likely to represent the population about which inferences are being made (Polgar & Thomas 1991). Powers et al (1995) conducted a study using the Elveru et al (1988) method to assess rearfoot posture in subjects with patellofemoral pain. The intratester reliability deemed excellent even though the tester was relatively inexperienced which the authors thought would prevent potential bias that could result from extended knowledge and expertise with this technique. Nevertheless, the tester was not blind to the subjects' identity that may limit the reliability of the results.

Elveru et al (1988) and Smith-Oricchio & Harris (1990) found that the reliability of therapists in positioning a rearfoot in STJN position is generally poor but can be moderately reliable if taken within-therapists than between-therapists over a short period. However, Elveru et al (1988) examined a group of inexperienced therapists after only 30 minutes of textbook instruction and Smith-Orrichio & Harris (1990) examined therapists with an unreported skill level. However, because of how this measurement is used and in light of the poor intertester reliability, the clinical usefulness of measuring the position of STJN is limited. Norkin & White (1995) elected not to use this position because use of STJN position adds error to ROM measurements. Picciano et al (1993) concluded that STJN measurements taken by inexperienced therapists are unreliable and recommend that clinicians determine their own reliability for this measurement. This can also be seen in research conducted by Pierrynowski et al (1996) which tested the proficiency of experienced foot care specialists (chiropodists) and untrained physiotherapy students to place the rearfoot at STJN position. It was readily apparent that the experienced therapists were able to locate the STJN position much better than the students were. However, in another similar study by the same authors in 1997, the STJN position was tested in three different positions-weight bearing, prone and seated and some of the experienced chiropodists found difficulty in finding STJN position due to the different positions of the patient. They showed that the ability of the foot care specialists to position seated subjects in STJN position was poor. Chiropodist number three, who was the most experienced foot care specialist, normally performed 80% of their STJN position assessments in a prone position and only 10% in the other two positions therefore did not have much more experience than the untrained students. Perhaps the reliability may be improved had the raters chosen a particular position that they were experienced with. Pierrynowski et al (1997) also agrees that it may be possible that a different patient position could improve the reported proficiency value and warrants further research.

2.5.5. Knee flexion and extension

Knee flexion and extension is the largest component of knee motion and occurs in the sagittal plane (McClay & Manal 1998).

Radiographic measurement of maximum knee flexion is considered the most accurate technique (Gogia et al 1987) but in clinical practice, most practitioners either visually estimate knee flexion ROM or measure knee flexion with a goniometer.

Gogia et al (1987) demonstrated high intertester reliability and validity for goniometric measurements at the knee joint and Rothstein et al (1983) found high intertester reliability values for flexion of the knee but poor intertester reliability for extension of the knee using three different goniometers. However, it was thought that this was a result of not having a standardised measuring technique as each therapist used a different patient position when measuring knee movements. Rothstein and colleagues tested this hypothesis and a *post hoc* analysis was carried out but even using the same standardised position between testers, the reliability was never comparable to intratester reliability standards. They concluded that for knee extension measurements, the same therapist should take all the measurements for one individual patient. Boone et al (1978) and Low (1976) also found higher intratester reliability than intertester reliability when measuring ROM at the knee joint and both agree it would seem to be better that where possible, one therapist should make all the measurements for a patient. In contrast, Rhealt et al (1988) found intertester reliability to be high when measuring knee movements with either a universal goniometer or a fluid-based goniometer however, only a small sample of subjects were used and all the subjects were healthy which does not represent an accurate patient population. In addition, only two testers were studied with only two recordings on each instrument. Perhaps more testers or making more measurements with each instrument would provide the study with more interesting and valuable information.

In a gait study conducted by Nadeau et al (1997), it was found that the knee flexion angle measured with a universal goniometer was significantly lower for the group with anterior knee pain as compared with the control group. However, Dillon et al (1983) hypothesised that anterior knee pain subjects used less flexion at the knee joint to decrease forces at the patellofemoral joint and thus avoid pain.

Taking all the past literature into account, many authors concur that using a Universal Goniometer can be more reliable when it is used by the same tester (Boone et al 1978; Grohmann 1983; Low 1976 and Rothstein et al 1983). Brosseau et al (2001) also found that it was more reliable than visually estimating the ranges of motion at the knee. This was the justification used for using this tool for the measurement of knee flexion and extension.

The American Academy of Orthopaedic Surgeons (1965) state that the normal knee flexion is 135 degrees whereas the American Medical Association (1990) states that it is 150 degrees.

2.5.6. Resting calcaneal stance position (RCSP)

The resting calcaneal stance position is considered to be an indication of the way the foot has compensated for various structural and functional abnormalities of the foot and lower limb (Payne & Richardson 2000). It is also an indicator of STJ motion when weight bearing. The clinical importance of measuring the rearfoot angle is related to the premise that excessive foot pronation is manifested by an everted calcaneus during standing (Cornwall & McPoil 2004). The everted rearfoot is generally thought to result from compensatory movements at the STJ and or midtarsal joints (Donatelli 1996).

It is important to note the difference between the RCSP and the STJN in order to assess the compensation that occurs for proximal and distal problems however there are limitations of comparing them in that they only provide information regarding frontal plane motion of the rearfoot (Merriman & Turner 2002). Some authors suggest that sagittal plane motion (Meuller et al 1993) or transverse plane motion (Nawoczenski et al 1995) may be a better indicator of foot pronation.

It has been proposed in the literature that the angle of the rearfoot in standing should be between 0° and 2° (Root et al 1977). However Sobel et al (1999) reported mean eversion values for adults of 6.07°.

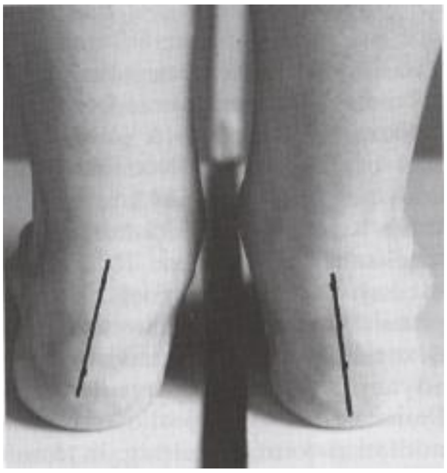


Figure 2.7 Feet in RCSP[6]

2.5.7. Rearfoot angle (varus/valgus)

The rearfoot angle is a measurement of the inclination of the calcaneus relative to the lower leg in resting position and is a typical component of the clinical examination of lower limb pathologies (Donatelli 1996).

The rearfoot angle which can sometimes be referred to as the calcaneal angle, was measured in relaxed single-limb stance as the angle between a line that bisected the calcaneus and a line that bisected the lower third of the leg. This was the same bisection line which was used to measure STJN and relaxed calcaneal stance position.

This measurement should be made under the assumption that in equal, relaxed standing, the STJ should be at or near its neutral position and indicates to the clinician what abnormal compensation may be occurring (Donatelli 1996).

[6] Merriman and Turner (2002)

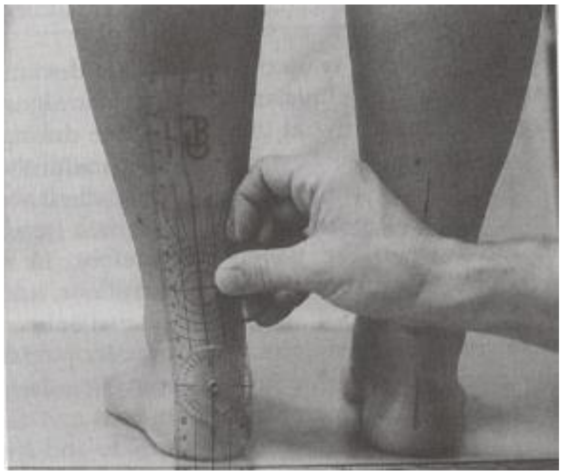


Figure 2.8 Measurement of rearfoot angle[7]

Despite the recurrent clinical use of the rearfoot angle, its direct relationship to abnormal rearfoot motion during walking and its relationship to lower extremity injury is sparse. Normal values of rearfoot angles vary somewhat throughout the published literature starting early with Root et al (1977) proposing that the rearfoot angle in relaxed stance should be 0±2°. However, other authors do not share these values. Sobel et al (1999) reported mean rearfoot values of 6.07 degrees of eversion in adults and Cornwall & McPoil (2004) had a finding of 6.3 degrees of rearfoot eversion. Donatelli et al (1999) described rearfoot angles of more than 8 degrees to be excessively pronated and angles between 1-8 degrees to be pronated. Root et al (1977) defined abnormal pronation as compensation at the STJ for a variety of lower extremity postures, resulting in excessive or prolonged eversion movement of the rearfoot during the stance phase of gait. They defined "excessive" as pronation movement greater than 4-6 degrees. Genova & Gross (2000) state excessive is equal to or more than 10 degrees and Eng & Pierrynowski (1994) and Johanson et al (1994) both report that anything more than 6 degrees of rearfoot motion during stance is excessive.

[7] Donatelli 1996

INVESTIGATOR	**CRITERIA FOR EXCESSIVE PRONATION**
Root et al (1977)	**Greater than 4-6 degrees**
Genova & Gross (2000)	**Equal to or more than 10 degrees**
Eng & Pierrynowski (1994)	**Greater than 6 degrees**
Johanson et al (1996)	**Greater than 6 degrees**

Table 2.1 Various authors' criteria for excessive pronation

In contrast to these studies, Cornwall & McPoil (2004) concluded that unless there is some pathology such as rheumatoid arthritis that introduces abnormal bone and joint structure and mechanics, all individuals pronate the same amount and that excess or overpronation is not due to rearfoot motion but a matter of exceeding the individuals soft tissue mechanical limits. They measured the rearfoot angle of 82 subjects and put them into groups of everted and inverted depending on their values. Rearfoot motion for the two groups was measured during walking and the kinematic data found no difference between the groups. Their findings indicated that a subject's rearfoot angle does not influence rearfoot motion during walking and therefore its usefulness is questionable.

2.5.8. 1st Metatarsalphalangeal joint (MTPJ) flexion

This joint plays an important and functional part of the gait cycle. Its general function is to allow the foot to progress through the terminal stance phase of the cycle. In 1954, Hicks published the first scientific examination of motion at the 1^{st} ray and reported that the 1^{st} MTPJ motion was controlled solely by the windlass effect of the plantar aponeurosis. The plantar aponeurosis is an extremely important structure in the maintenance of the longitudinal arch of the foot and absorbing stress for weight bearing. Donatelli (1985) states "*in the static stance position the plantar aponeurosis takes up approximately 60% of the stress of weight bearing and the beam action of the metatarsals approximately 25%.*" Hicks (1954) understood that since the plantar aponeurosis is attached proximally to the medial tubercle of the calcaneus and distally to the proximal phalanges of the toes through its connection to the plantar pad, then when the hallux is dorsiflexed or extended, the plantar pads and the plantar aponeurosis will move distally around the heads of the metatarsals. Hicks (1954) also stated "*the effect was as though a cable had been wound one quarter of a turn on the drum of a windlass; the drum of the windlass being the head of the metatarsal, the handle which does the winding being the proximal phalanx, and the cable which is wound on to the drum being the plantar pad and the plantar aponeurosis.*" This effect will cause the rearfoot to supinate as the pull of the plantar aponeurosis causes the distance between the calcaneus and the head of the 1st metatarsal to shorten therefore increasing the height and length of the arch. He tested this method on a living foot and a cadaver specimen and found that all the effects were present on both subjects therefore concluding that the "windlass mechanism" operates independently of muscular activity (Aquino & Payne 1999).

The ability of the hallux to undergo dorsiflexion in gait has been proposed as an essential element of normal locomotion (Harradine & Bevan 2000). Dananberg (1993) states that normal ambulation requires at least 65 degrees of motion. Restriction of motion of this joint can severely mar the function of the foot and altered gait patterns and pathological changes in the joint may occur consequently. Hallux limitus is a foot deformity often seen associated with a limitation of dorsiflexion in the 1st MTP joint. Clinical assessment can be conducted by observing the decreased ROM of the joint. While normal dorsiflexion ranges from 50 to 70 degrees, hallux limitus involves less than 50 degrees of motion (Birke et al 1995). Patients will complain mostly of pain or crepitus in the joint and symptoms are usually insidious and progressive. Treatment can be successful if managed in the early stages but may require an orthotic device to help the patient, as there is insufficient ROM in the joint to support the hallux in the toe-off stage of the gait cycle.

Harradine & Beran (2000) conducted a study to test if there was a decrease in hallux dorsiflexion with rearfoot pronation. The results showed that there was an initial increase in the ROM of dorsiflexion with an increase of pronation but only up to a point of five degrees eversion. Any additional degree of pronation showed no real significant decrease in 1st MTP joint dorsiflexion. However, these results have restricted integrity, as the sample size was very small it was therefore not representative of a normal population and the study did not consider dynamic structure only static. Further research is needed to determine if limitation in 1st MTP joint ROM is sufficient to cause lower extremity injuries and if so, how much limitation is needed before an injury occurs.

The reported values of dorsiflexion of the 1st MTP joint are very variable according to (Oatis 1988). Oatis (1988) stated dorsiflexion of the great toe to vary from 0-70 to 0-90 degrees and Merriman & Tollafield (1995) found the same for dorsiflexion. They report that normal dorsiflexion of the first MPT joint should be 70 degrees as does The American Academy of Orthopaedic Surgeons (1965) but the American Medical Association (1995) state that normal dorsiflexion should be 50 degrees.

Goniometers are widely used in clinical practice and have been demonstrated as a reliable and valid instrument when measuring 1st MTP joint dorsiflexion (Roukis et al 1996). Hogan & Kidd (2001) conducted a single-blinded randomised study of thirty subjects with pain of the 1st MTP joint. The study assessed the validity and reliability of using a universal goniometer to measure ROM of the 1st MTP joint. The results suggested that it is reliable and valid when used by the same therapist on the same subject however, when the angle was being recorded, the value was rounded to the nearest whole degree as the goniometer only had measurements in whole degrees. This resulted in measurement error but because there is no other reliable, valid, and affordable technique yet available, the universal goniometer remains a popular choice for measuring the 1st MTP joint (Hogan & Kidd 2001). The flexible electrogoniometer has also been reported to be highly accurate and reliable in the measurement of this joint (Ball & Johnson 1993).

Hopson et al (1995) investigated the intrarater reliability of four methods of measuring the first metatarsalphalangeal joint and found that although the mean measurements for all four methods varied considerably, the measurements obtained by each method were reliable. They also stated that although each of the measurement methods was reliable, they should not be used interchangeably.

2.5.9. Arch height

The arches of the foot became important structures of the human body thousands of years ago when man's ancestors stood erect and began bipedal locomotion (Saltzman & Charles 1995). The plantar aspect of the foot is divided into two arches, the transverse arch, and the longitudinal arch. The transverse arch is supported by the three cuneiforms and the cuboid bones. The longitudinal arch is divided into two parts: a medial longitudinal arch (MLA), which supports the medial aspect of the foot and a lateral longitudinal arch, which supports the lateral aspect of the foot. The MLA is supported by the calcaneus, talus, navicular, medial, intermediate, and lateral cuneiforms and the first and second metatarsals.

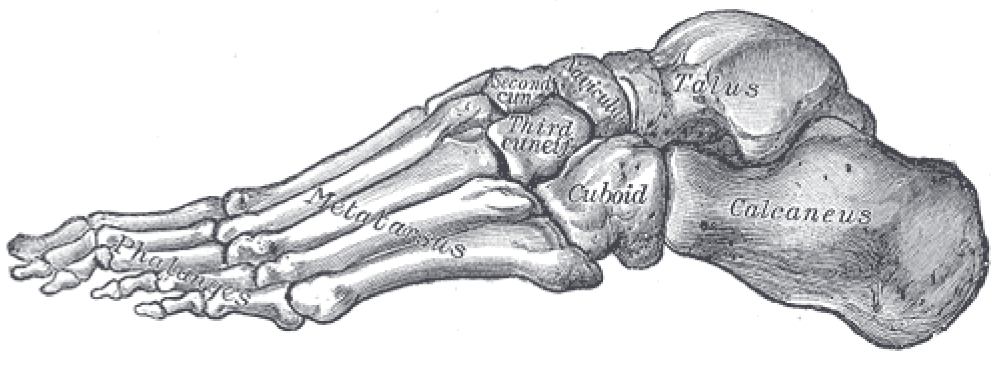


Figure 2.9 Bones of the medial longitudinal arch[8]

[8] Williams et al 1989

The lateral longitudinal arch (LLA) consists of the calcaneus, cuboid and the fourth and fifth metatarsal bones. Ligaments, muscles, and a deep fascia on the plantar aspect of the foot called the plantar aponeurosis support the arches. The longitudinal arches play an important role in man's ability to keep upright and balanced and in shock absorption. The arch in connection with the soft tissue heel pad acts as a shock absorption during running, walking and jumping to absorb some of the impact forces from the ground (Saltzman & Charles 1995). The shock absorber mechanism is carried out by the combined efforts of the thick soft tissue heel pad, the flexible joints of the longitudinal arch, the plantar ligaments and the plantar aponeurosis (Manusov 1996).

The height of the sagittal plane arch (MLA) has been one of the primary criteria for classification of foot structures (Saltzman et al 1995) and can be determined by the shapes of the bones and the laxity of the ligaments of the foot (Harris & Beath 1948). It is measured from the highest point of the soft-tissue margin of the medial longitudinal arch, which, anatomically would be the navicular, to the ground and should be a relatively easy procedure to conduct. However, it can be confounded by bony architecture and soft-tissue variations between subjects and there is yet no existing objective criterion for the classification of foot type's pes cavus and pes planus. Pes Planus is a complication of the foot in which the height of the MLA is lower than normal or even absent. An individual with pes planus may be referred to as having flat feet (Manusov 1996). Pes cavus is a complication of the foot in which the height of the MLA is greater than normal. Both of these conditions greatly reduce the foot's ability to act as a shock absorber for the rest of the body.

The height of the MLA is commonly thought to be a predisposing factor to knee injuries (Williams & McClay 2000 and Kaufman 1999) although there are many opinions in the literature regarding this. Some of the controversy may be due to the many different ways of measuring the MLA but there is however, no unanimous stand on how this evaluation should be carried out (Sneyers et al 1995). There are to date, no universally accepted clinical or radiographic definitions of the average height or the normal range of heights of the MLA. The point at which a low arch becomes a flat foot is unknown (Mosca 1995).

Although a variety of methods have been used in order to measure arch height such as with the use of X-rays, ultrasound, radiographs (Cobey & Sella 1981) and footprint parameters, these methods are limited as X-rays and ultrasound both require other specialised therapists to carry these tests out and are also expensive. X-rays, in addition, imply a potential health risk (Hawes et al 1992). In order to conduct a complete clinical evaluation of a subject's feet whether they are normal, flat or high arched, a detailed history, physical examination, and possible radiographic examinations are required. This is also time consuming and impractical (Cowan et al 1994).

Weight- bearing and non-weight- bearing examinations of feet can have very different outcomes according to Subotnick (1985) and Williams & McClay (2000). Establishing reliability in both weight-bearing and non-weight- bearing conditions allows for measurements that can be taken under both conditions and, therefore, may be used to describe foot mobility as both foot structure and foot mobility may play an important role in predicting injuries (Williams & McClay 2000). Although most of the methods attempt to quantify the arch, some methods are based on observation. Giladi et al (1985) classified the non-weight -bearing foot as either high-arched or low-arched by visual assessment alone. Even among very experienced clinicians however, this method is highly inconsistent (Cowan et al 1994). In the study conducted by Cowan et al (1994), they found that there was poor agreement between clinicians in visually assessing arch height based on observation of photographs. Among those at the extremes of the evaluations, the high arch foot was predicted much less reliably than the flat foot thus concluding an unacceptable level of intertester variability. This study strongly argues for a standardised quantitative measurement of arch height at least as a screening mechanism.

Harves et al (1992) measured the highest point of the soft tissue along the MLA in full weight-bearing, however Williams & McClay (2000) do not believe that this measurement necessarily represents the state of the bony architecture of the foot. Dr. M. Kvist of the Sports Medical research Unit at the Jyvaskyla University in Turku developed a method for measuring arch height. The foot arch was evaluated while the subject was standing on a podoscope with feet in slight abduction. A mirror, mounted underneath the glass plate, allows the therapist to evaluate the plantar contact surface with the glass plate. When the arch is low, the contact surface in the midfoot region is often wide, when the arch is high the midfoot region contact barely touches the ground and normal arches usually have intermediate midfoot contact surfaces. However, although Sneyers et al (1995) used this method they concluded that having only three subdivisions of arch heights was too general and larger divisions need to be tested.

Various authors have studied anthropometrical techniques of measuring arch height. Williams & McClay (2000) measured the reliability and validity of seven measurements of the arch compared to radiographically determined bony landmarks and found only two out of the seven measurements to be similar to the radiographs. Nigg et al (1993) and Hawes et al (1992) both emphasise the indistinctions and inconsistencies of measuring the MLA by non-invasive means. They found that when measuring arch height on healthy subjects, anthropometric techniques seemed to be reliable but the validity related to radiographs was never tested. Saltzman et al (1995) found that clinical or anthropometric measurements yield reliable and valid approximations of the MLA structure when certain criteria are met. Experienced examiners record the measurements, the same measuring device is used across testing conditions, and when measurements are taken in a weight-bearing position are such examples which have been reported to allow for highest reliability estimates (Elveru et al 1988 and Smith-Oricchio & Harris 1990).

The most popular approach has involved the interpretation of footprints. The footprint as a measure of pathological conditions has been in use for some time. They are simple to obtain, inexpensive and if properly taken they show a strong visual observation of the height of the longitudinal arch (Clarke 1933). Footprints may be made with home-made devices such as a solution which is brushed over the sole of the subject's foot and the foot is then placed with full weight on a piece of paper. Several commercial organisations also provide materials for footprints. The pedograph machine is a popular device as it is inexpensive and easy to use (Clarke 1933). Although footprints are reliable (Clarke 1933), their validity is suspect (Saltzman et al 1995). Validity may be determined by computing measures with other evidences of the equalities they are to measure (Clarke 1933). The ability of the simple footprint however to predict dynamic rearfoot motion remains questionable. Atkinson-Smith & Betts (1992) conducted research in which simple ink footprints and dynamic pedobaragraphy were compared to see if the ink prints were useful indicators of dynamic rearfoot motion. The authors concluded that the ink print did not provide any information regarding plantar foot pressure throughout the gait cycle and were therefore deemed invalid in the measurement of dynamic motion. Footprints, obtained from the Harris-Beath mat, have also been used to describe foot architecture (Hawes et al 1992 and Thomson 1994). Hawes et al (1992) found that they were unable to predict arch height from the footprints and concluded that they were indicators of footprint shape only. Song et al (1996) also agree and believe that although these mats provide a reasonable method to describe the gross foot morphology such as low arch, high arch, or normal arch, they do not quantify the specific forefoot to rearfoot alignment. William and Morrison (1931) doubt the ability of the footprint to indicate foot type since factors other than arch height may contribute. Williams & McClay (2000) also doubt the ability of footprints, as the soft tissue on the plantar aspect of the foot is thick and variable and can mask the true bony architecture of the foot. Urry & Wearing (2001) compared electronic footprints against ink footprints to determine the relative accuracy of the area and angular measurements of the electronic footprint. A Musgrave Footprint foot pressure platform was used to collect the electronic measurements and a foam rubber pad impregnated with water-soluble ink was used to collect the ink footprint. The study revealed that the pressure platform consistently underestimated the contact area of the foot by as much as 14% and the electronic footprint produced a small yet significant reduction when calculating the arch index. Consequently, the accuracy of measurements obtained from electronic prints is questionable and are not representative of those footprints derived from an inkpad.

When measuring the height of the longitudinal arch, Saltzman et al (1995) considers radiographic measures as "gold standard" in their study due to the reliability of radiographic studies using experienced examiners to be consistently high and also, the radiographs provide a clear-cut image of the skeletal components of the MLA.

It has been suggested that a functional relationship between arch height and knee injury may exist. It is generally understood that a low arched foot tends to be more flexible and therefore is subject to increased pronation during the contact phase of the gait cycle and in contrast, a high arched foot is more rigid and subject to increased supination. Consequently a low or high arched foot may place the athlete at a higher risk of injury (Razeghi & Batt 2000). Lutter (1980) states that the high-arched, cavus foot has been associated with knee pain. However Cowan et al (1993) proved in their study that low arched feet were more prone to injury than high arched feet. They concluded that a low arched foot actually provides protection against lower limb injury. However, another study conducted by Duffey et al (2000) of normal subjects and subjects with AKP, found the AKP group had a significantly lower arch height relative to the control group, which indicates that a higher arch foot type is preventative of AKP.

McKenzie et al (1985) stated that the configuration of the MLA is a valuable method of classifying feet and has direct implications on the development and management of running problems. Nigg et al (1993) demonstrated that the transfer of foot inversion to internal leg rotation was found to increase significantly with increasing arch height. It was suggested that this relationship might explain the cause of knee pain. Saltzman et al (1995) also thought that there is a functional relationship between arch height and injury and it may be related to a transfer or "coupling" motion between the leg and foot. Kaufman et al (1999) however does not support the association between general foot types and overuse injuries at the knee.

2.5.10. Navicular height/drop

The drop of the navicular tuberosity in relation to the bony landmarks within the foot is a further test to be a measure of STJ pronation and excessive MLA collapse (Gross 1995). Brody (1982) was one of the first authors to report navicular drop as a measure of foot pronation in runners. Sell et al (1994) evaluated the reliability of navicular drop and found high intratester and intertester reliability. They concluded that this method is a far favourable and more reliable method of measuring foot pronation and is much easier and simpler to perform than the traditional method of bisecting the calcaneus. Mueller et al (1993) also agreed with this method and stated that navicular drop can be considered a reliable "composite" measure of foot pronation. In contrast, however, Picciano et al (1993) reported differing results with poor-to-fair intratester reliability and poor intertester reliability. The conflicting results may be due to the fact that the testers used were very inexperienced physiotherapy students both of whom had only two hours to practice the method and as mentioned before by the authors and by Pierrynowski et al (1996), experienced testers has been proven to improve the reliability of clinical measurements. Had the inexperienced physiotherapy students had more time to practice or they had used experienced physiotherapists, perhaps the results of the study may have been in agreement with Mueller et al (1993) and Sell et al (1994).

There are many differing methods to measure navicular drop and Menz (1998) measured navicular drop by placing the fully weight- bearing subject in the talar head congruent position (STJN position) and measured the distance between the navicular tuberosity and the supporting surface. The subject is then told to relax, and the sagittal plane excursion of the navicular is measured with a ruler. When describing this method however, Menz (1998) gave no indication of the sample group or evaluated the reliability of this method.

Beckett et al (1992) described values as 6mm or 7mm up to 10mm but it is not clear how these figures were obtained. Brody (1982) suggested that normal navicular drop is approximately 10mm and that measurements over 15mm represents abnormal pronation. However, both of these authors fail to take into account the size of the foot being assessed. A navicular drop of 15mm may be excessive according to Brody (1982) on a small foot but may be normal drop on a larger foot. Further research is obviously needed to establish a navicular drop "index", that is, the amount of navicular drop considered normal or abnormal relative to the size of an individual foot (Menz 1998).

The reliability of this measure has been questioned (Picciano et al 1993) however. Williams & McClay (2000) conducted a study, which tested the reliability, and validity of the height of the navicular based on their use of bony anatomical landmarks. It was measured from the floor to the most anterior-inferior portion of the navicular. To establish concurrent validity, they compared the clinical measurement with radiographs. They found that the mean values for navicular height were found to be in agreement with values from other studies such as Cavanagh et al (1997), Saltzman et al (1995) and Wen et al (1997). However, the absolute values for navicular height was much lower than previous studies such as Cowan et al (1993), but it was reported that Cowan and colleagues took their measurements from photographs which could account for the difference in values. Williams & McClay (2000) concluded that intratester reliability was much higher than intertester reliability when measuring navicular height. Saltzman et al (1995) found good intratester reliability and moderate intertester reliability when measuring navicular height. Sell et al (1994) also found excellent intratester and moderate intertester reliability. Weiner-Ogilvie & Rome (1998) calculated navicular height as the most prominent palpable portion of the navicular tuberosity was marked with a dot using a fine black pen when the subject was in a prone position. Callipers were used to measure the distance from the fixed end of the calliper to the mark on the navicular tuberosity. They found that intratester reliability was much higher than intertester reliability by a large degree. This finding was in agreement of other studies such as McPoil & Cornwall (1996) and Smith-Oricchio & Harris (1990) however, it should be noted that the sample size in the former study was small and therefore the population choice was limited. McCrory et al (1997) suggested that radiographic measurements could reliably determine the height of the navicular. By measuring with this method, the errors involved in palpation, skin movement and soft-tissue distribution are minimised. Thompson (1994) concluded that radiographs also exhibit significant intratester and intertester variability.

2.6. A Review of the Literature on the Vicon 370 Kinematic Motion Analysis System

2.6.1. Introduction

Computerised three dimensional (3D) gait analysis has been increasingly used for quantifying gait analysis and therapeutic intervention however it will only receive wider acceptance in clinical practice if its reliability can be demonstrated (Maynard et al 2003). It is essential that the measurements taken can be repeated by the same tester on different occasions and yield similar results especially when clinical inferences are drawn. Winter (1984) conducted a study reporting on intrasubject repeatability of kinematic and kinetic data on two subjects where one subject was evaluated on different test days and the other was evaluated on the same day but different times. The results showed that intrasubject repeatability was better within the same day of testing compared to results from different days.

However, there are many sources of error associated with 3D gait analysis. One commonly recognised problem is the day-to-day variability that may be present due to marker placement on the skin. This is especially important when the same subject is being tested on more than one occasion and comparisons are to be made between the sessions. Skin movement, mathematical model assumptions and anatomical marker placement are also common sources of error. Accuracy of marker based systems is affected by errors introduced due to skin movement (Benoit et al 2006). The results may not reflect the motion of the underlying bones and skin movement was not quantified in this thesis. However, any errors relating to skin movement would be equally apparent in all subjects and would therefore produce no bias. The results must be interpreted with this in mind.

2.6.2. Reliability and Validity of Vicon 370 Kinematic Motion Analysis System.

Measurements of the spatial and temporal parameters of gait patterns are frequently obtained to identify gait deviations, aid with diagnosis and to determine and monitor patient progress (Bilney et al 2003). Making evidence-based decisions in the clinical management of gait dysfunction requires the ability to measure temporal-spatial gait variables validly and reliably (Barker et al 2006). There have been several significant studies which have investigated the validity and reliability of various motion analysis systems but the reported findings have not been consistent.

Reliability of a measurement tool incorporates repeatability and precision of its measurements (Durward et al 1999). Identifying the location of any anatomical landmark by palpation or other means in a repeatable fashion is of crucial importance to the reliability of relevant results in any experimental approach. Careful attention should be paid to sources of error, mainly due to the relative displacement between marker arrays associated with skin and underlying bones, which affect the determination of instantaneous positions and orientations of lower limb bones (Benedetti et al 1998).

Face validity refers to the judgement made about the appropriateness of the measurement tool for its intended use (Durward et al 1999). It has been demonstrated by the reported ability of Vicon to measure all the parameters of interest in this particular study and by its use in previous studies (Ferber et al 2002 and Williams et al 2003).

Content Validity refers to the particular measurement tools ability to measure all aspects of the behaviour of interest (Durward et al 1999). It was demonstrated by the ability of the Kinematic System to measure all gait and relevant parameters.

Kadaba et al (1989) investigated the repeatability of gait variables using Vicon with 40 normal subjects, three times daily and on three different test days. They used a statistic which measured the overall similarity of waveforms taking into account the concurrent effects of differences in offset, correlation and gain called the coefficient of multiple correlation (CMC). They found excellent intratester repeatability of joint angle motion at the hip, knee and ankle in the sagittal plane within a test day and good repeatability between test days within (CMC = 0.643-0.996) and between (CMC= 0.240-0.944). Pelvic tilt pattern displayed the lowest repeatability (CMC= 0.598) for within test days and (CMC= 0.529) for between test days. The authors stated the reason for the low values could be due to the pelvis's small range of motion (mean range 1-2 degrees). The results for the joint angle motion in the other two planes were found to be reduced than those in the sagittal plane and repeatability was much better within a test day than between test days. This, however, is to be expected as within test day repeatability results are not affected by marker re-application errors.

Kadaba and his colleagues write that the reason for the excellent repeatability only in the sagittal plane is due to the fact that a higher level of control is exercised by the neuromuscular system since the direction of progression is along the sagittal plane. They reinforce this by stating that the results of the between test days in the sagittal plane are also excellent. In conclusion, they suggested that the results demonstrate that the gait variables are quite repeatable and it would be reasonable to base significant clinical decisions on a single gait evaluation using the Vicon Kinematic Motion Analysis System.

Another study assessed the reliability of gait measurements and resulted in different findings from the above study. Maynard et al (2002) investigated the intratester and intertester reliability of kinematic data using the dual CODA mpx30 (Charnwood Dynamics, Barrow on Soar, Leicestershire, England) Motion Analysis System. Interestingly, they found better intertester than intratester repeatability for most of the gait parameters measured. Test-retest repeatability of measurements of all joint kinematics was best for the knee angles and poorest for the hip angles. This is consistent with observations from a previous study by Cowman et al (1998) and may be due to the easier identification of the anatomical landmarks for the placement of the markers on the knee. Both studies findings do not demonstrate complete reproducibility of the gait analysis data when measurements are made with the CODA mpx30 System.

Barker et al (2006) compared the accuracy and reliability of measurements taken with the Gait Mat II System (GM) with the same measures taken simultaneously with the Vicon Kinematic Motion Analysis System. The GM is a portable and inexpensive device designed to collect spatiotemporal characteristics of gait. The data from the GM can be analysed to provide the same common gait parameters as the Vicon System can. Measurements taken simultaneously by the GM and Vicon had an ICC of (0.99) indicating excellent reliability. The results support the reliability and concurrent validity of the GM as a clinical gait analysis tool (Barker et al 2006). This is particularly useful as although Vicon is known as the "Gold standard", it has its disadvantages. It is not portable and convenient, expensive and it is not easy to use in a routine clinical environment.

The Vicon 370 Kinematic Motion Analysis System has been shown to be one of the most accurate optical measurement systems in a comparison by Ehara et al (1997) with a reported mean absolute error of (0.94mm). For this reason, Vicon is often considered the Gold standard in motion analysis (Ehara et al 1997).

2.6.3. Marker placement

The Helen Hayes marker set is a relatively simple set of external markers developed for time-efficient video analysis of lower extremity kinematics. The original configuration of 13 markers was developed by Kadaba et al (1988) at the Orthopaedic Engineering and Research Centre, Helen Hayes Hospital, West Haverstraw, N.Y. The basic Helen Hayes marker set consists of 15 lower body markers and the markers are secured to the body at anatomical significant locations that determine embedded axes for segments under consideration. The Helen Hayes marker sets determine ankle and knee joint centers and segment coordinate systems by means of a marker on a post or wand protruding from the lateral aspect of the thigh and shank, and by single markers placed over the lateral aspect of the joint flexion/extension axis. It is important that the wands do not move with respect to their original position on the patient's legs during data capture. If the wands move, the knee and or ankle coordinate systems will move with them and undetectable movement can cause large degrees of inaccuracy in joint kinematics and kinetics.

Intratester reliability of marker placement is very important to establish especially when reliability is in part dependent on placement of markers by the operator. If the skin above the bony landmark is liable to displacement, marking should be carried out with these bones in the relative position they assume when the anatomical landmark calibration procedure is carried out. For example, ASIS landmarks may be displaced if they are marked when the patient is supine and they are slightly overweight to when they stand up.

Despite this model being a very popular and common marker set, it does have its limitations when measuring the foot. Measurement of foot kinematics is becoming increasingly popular as motion analysis systems become more and more accurate. However, an accurate measurement of the movement of the talus during walking is impossible using non-invasive techniques since there are no external anatomical landmarks are present for marker placement. As a result, there is inadequate information on the function of the rearfoot (the talus and the calcaneus) during the stance phase of walking. Therefore a valid and repeatable multisegmental foot model is needed for understanding normal and pathological function, planning intervention and evaluating the outcome of treatment (Stebbins et al 2005).

2.7. Review of the literature available on past and present foot models.

Over the last few years, various authors have presented in vivo studies of the foot and ankle complex on healthy adults (Carson et al 2001; Kidder et al 1996; Leardini et al 1999; Hunt et al 2001; MacWilliams et al 2003) but quantitive comparisons between them are not possible due to the differences in marker placement and definitions of fixed anatomical axes (Leardini et al 1999). There is a need for a standardised multi-segmented foot model and measurement protocol applicable to gait analysis. There are numerous ways of measuring foot mechanics such as cadavers, surface markers, bone pins and markers and imaging. Each has its own set of advantages and disadvantages associated with them. One of the key issues to consider when implementing a kinematic foot model is the potential for skin artefact errors (Reinschmidt et al 1997). They tried to determine the effect of skin movement artefact on the calculation of tibiofemoral motion during running using intracortical Hofmann bone pins with reflective markers attached. They found that the agreement between the skin and bone markers was generally poor for abduction/adduction and internal/external knee rotation motions. Conversely, skin mounted markers were able to give a good representation of flexion/extension of the knee and was in agreement with the results for walking and the skin movement errors were consistently higher for running compared to walking (Reinschmidt et al 1997b).

The designation of anatomical coordinate systems (Leardini et al 2007) and the definition of a "neutral" position of the joint of the foot (Leardini et al 2007; Liu et al 1997) are also issues to consider when designing a biomechanical model.

Leardini et al (1999) proposed an in-vivo technique using five rigid segments. Each segment was assumed rigid and was identified with an anatomically based co-ordinate system. They used rigid arrays of markers as they claimed this embraced the underlying bones better than skin mounted markers. They were placed on the foot and ankle according to specific locations of anatomical landmarks using the tip of a pointer which is more accurate and practical than direct marker placement especially when they are located in awkward positions such as the head of a metatarsal or a pointed tip of malleoli. However, Carson et al (2001) also developed a foot model for healthy adults using three segments which was non invasive and very similar to that of Kidder et al (1996). They used skin mounted markers which saved time required for the rigid array of markers used by Leardini and workers (1999) in landmark identification. Carson et al (2001) also mentioned skin motion artefact which affects the array of markers in a uniform manner and is therefore impossible to filter mathematically.

When human movement is measured using motion analysis, each marker attached on the body surface moves together with the underlying skin, which during movement moves with respect to the underlying bone. The amount of skin deformation depends on the physical characteristics of the subject, the location of the marker and the phase of movement being performed (Croce et al 2005). The relative movement between marker and bone represents a soft tissue artefact which affects the estimation of the skeletal segment and joint kinematics and is regarded as the most critical source of error in motion analysis.

Benoit et al (2006) conducted an investigation to quantify the error caused by skin movement artefact when reporting the kinematics of the tibiofemoral joint during movements which incorporate sagittal and non-sagittal plane movements. They hypothesised that skin movement error will reduce the ability to accurately measure 3D kinematics and that non-sagittal plane movements will be most affected by skin movement artefacts. The study indicated that skin mounted reflective markers display significant limitations in predicting 3D kinematics of the knee joint. Holden et al (1997) and Reinschmidt et al (1997) performed studies in which steel pins were inserted in the bones of volunteers and the positions of skin markers compared with the position of markers on the pins were analysed. Both results showed that the amount of movement depended on which parameter was actually being measured. Whittle (2002) agrees with these authors and states that marker movement has little effect on sagittal plane angles because it causes only a small relative change in the length of fairly long segments.

A reasonably new development within 3D analysis is the verification of a multi-segment foot model which differentiates between the forefoot and the rearfoot. The model is called the Oxford Foot Model and was defined by Carson et al (2001).

Figure 2.10 Oxford foot model[9]

It is based on two main segments, the rearfoot and the forefoot and an additional hallux segment and produces adjusted kinematics for the ankle as well as the added inter-segment angles. Limited information is available regarding the repeatability and error of this model in adults and knowledge of these would enhance the ability to interpret both individual differences (pre- and post intervention) and group differences (asymptomatic and symptomatic individuals) during kinematic investigations (Wright et al 2010). Stebbins et al (2006) and Carson et al (2001) published studies which both comment on the good repeatability of the Oxford model in healthy adults however there are very few published studies examining the inter-centre repeatability or validity of the model (Curtis et al 2009). Wright et al (2010) found that rearfoot and forefoot motion displayed high reliability and low error of adult gait. They used a slightly modified version of the Oxford foot model which improves the repeatability of the model according to Stebbins et al (2006). More research is required to provide valuable insight into rearfoot, midfoot and forefoot pathomechanics.

The last section of this literature review chapter introduces and reviews the measurement tool used in chapter 7 which is the Positional Upright MRI Scanner.

2.8. Magnetic Resonance Imaging (MRI) – A literature review

Various authors have conducted research to test the reliability of radiographic measurements albeit using plain X-rays or supine MRI's but there are very few

[9] Vicon motion systems (2011)

studies conducted on the lower leg using the Upright Positional MRI Scanner. This makes it unfeasible for comparison purposes; however there are a few papers available which have examined the spine. Gilbert et al (2008) stated that imaging the spine in the upright or weight-bearing position may increase the diagnostic accuracy for the medical professional. This accuracy may be further enhanced by placing the patient upright in the position that causes pain.

Supine MRI findings can often correlate poorly with clinical findings but upright MRI in the flexed, extended, rotated and standing positions allows patients to reproduce the positions that evoke their symptoms and may uncover MRI findings that were not visible with routine supine imaging (Alyas et al 2008).

Lin et al (2004) found that the radiographic approach is generally considered as the clinical gold standard when describing the medial longitudinal arch in the foot because it provides consistent reliability and a strong correlation between radiographic parameters and injury of the lower limb. They also state that whilst it has good reliability if performed by an experienced technician, it has the disadvantages of being time-consuming, costly and most importantly radiation exposure.

Normal reference range values for weight-bearing foot radiologic measures have been established by various authors with good agreement (Gentilli et al 1996) however the reliability and repeatability of foot radiographs has been investigated with varying results (Bryant et al 2000). Bryant et al (2000) found high ICC's for navicular height (0.92) and CIA (0.87). These measurements were conducted on radiographs and followed a standardised radiographic measurement technique. Menz and Munteanu (2005) showed that the test-retest reliability of each of their radiographic measurements was excellent with all the ICC values greater than (0.98) and the lower 95% CI consistently greater than (0.95). This level of reliability is consistent with previous reports.

Saltzman et al (1995) showed that a high reliability can be obtained from selected measurements when using the same tester. They correlated measurements obtained from radiographs taken at 50% of weight-bearing. The high established intrarater reliability values from navicular height (ICC=0.92) arch height (ICC=0.91) and calcaneal inclination angle (ICC=0.99) demonstrates that measurements based on clearly defined anatomical landmarks can be reliably measured from radiographs. However, this study reported that the measurements taken by the same tester within the same day were the most repeatable with a measurement error of up to 2 degrees but when using different testers, a measurement error in excess of 5 degrees occurred.

Murley et al (2009), Nawoczenski et al (1998) and Menz & Munteanu (2005) also found "gold standard" values for cancaneal inclination angle (CIA), (ICC=0.98, 0.97 and > 0.98) respectively when measuring the CIA radiographically.

Lohrer et al (2008) conducted a study measuring talar tilt at the ankle on 33 bilateral radiographs and found high values (ICC=0.83) for intratester reliability and unusually higher results for intertester reliability (ICC=0.95). These angles were measured manually with a goniometer and a ruler and measurements were made to the nearest 1 degree. They also measured NH (navicular height) from radiographs and clinically and they found clinical measurements of NH very strongly associated with the corresponding NH measurements obtained from radiographs (r values ranging from 0.72 to 0.76). Williams & McClay (2000) found clinically determined navicular height (r = 0.91) to be strongly correlated with radiographic navicular height.

Sanfridsson et al in 1998, tried to test a computed system for measuring radiological angles as most musculoskeletal measurements in conventional radiography are made with a ruler and a protractor directly onto the film. They looked at measuring medial and lateral joint spaces while testing intra-observer variations. The results were good and the standard deviations (SD's) were small. They concluded that although the use of measuring assistance tools for evaluation are time-consuming, they are important for the implementation of computed radiography in workstations dedicated to musculoskeletal radiography.

2.9. Conclusion

The main aim of this study is to investigate any association between lower limb biomechanics in subjects with anterior knee pain using the Upright Positional MRI

Scanner and the Vicon 370 Kinematic Motion Analysis System when barefoot, shod and wearing orthoses. In order to facilitate this objective, comprehensive background and literature reviews had to be conducted to understand which research had already been performed, protocols and methodology of measurements, statistical analysis of results and conclusions. This identified that there are very few consistent research papers available which can be compared in the field of lower limb biomechanics and foot orthoses. There appears to be lots of controversy between authors on whether or not orthoses actually do affect the biomechanics of the foot and leg. This is due to differing methodology, different types of orthoses and various measurement tools utilised. There is also a lot of speculation on the kinematics of the rearfoot due to the fact that previous foot models only identified the foot as a whole segment and not as having a midfoot, rearfoot and a forefoot segment. This however should change due to the design of the Oxford foot model which is constantly being updated in order to improve repeatability and remove immediate sources of error.

Repeatability and sources of error were discussed as these are paramount to any source of research. Reliability and validity of measurement is a fundamental part of clinical practice, particularly when clinical assessment is based on subjective judgements for diagnosis, choice of potential intervention, and a review of management (Keenan & Bach 1996).

It is also paramount that in order to produce reliable and repeatable results, the actual clinical protocol for each measurement requires to be well documented and known for its reliability and validity by previous authors. This should generate more consistent results which can be comparable to previous studies.

The two main measurement tools used in this study are the Vicon Kinematic Motion Analysis System and the Upright Positional MRI Scanner. These have been explained in depth and limitations and restrictions which may occur have been duly noted. Examples of these are marker placement and skin movement with the kinematic analysis and “ghosting” and the partial volume effect with the radiographs.

Chapter 3 – Intrarater Reliability of Clinical Assessment

3.1. Introduction

In order for a clinical measurement to have any scientific credibility, a reliable and valid clinical measurement technique must be employed. The aim of this chapter is to develop a standardised and reliable protocol for a clinical assessment by measuring the tester's reliability (intratester) of performing the measurements required. With regard to clinical assessment, reliability can be defined as the amount of agreement between successive measurements of the same joint by the same tester or different testers, namely, intratester and intertester reliability respectively. This is an important pre-requisite for the protocol so that the same clinical assessment can be applied to both the analysis of asymptomatic and symptomatic subjects in the main study.

3.2. Methodology of reliability study

This experimental study took place at the Grampian Gait and Motion Analysis Centre in Aberdeen. Full ethical approval from the local research and ethics committee was received prior to data collection.

Subjects in this study were required to meet the following inclusion/exclusion criteria prior to data collection. This was based on those used in other previous reliability studies (Bennell et al 2000; Klingman et al 1997). Inclusion criteria included initial screening determined all subjects to be free of pain, musculoskeletal and neurological dysfunction of both right and left lower extremity. Exclusion criteria included that there was no intervention of foot orthoses, none of the subjects had been treated for musculoskeletal disorders of the right or left lower extremity in the past and none of the subjects were qualified physiotherapists or podiatrists.

All volunteers were recruited from the RGU student population and were familiar with the purpose of the study and signed a consent form prior to participation. It should be noted that all of the subjects were healthy with no apparent injuries and, thus, generalisation of the results to a patient population may be restricted. However, as the aim of this pilot study was only to test intratester reliability of a random sample of clinical measurements before conducting a larger study, it was felt that the healthy sample used would be deemed appropriate.

One tester (the researcher) with experience in measuring joint ROM carried out 15 goniometric measurements and one tape measurement on 5 healthy volunteers ranging from ages 31-45 years. The same tester performed fifteen different measurements on each subject in three different measurement sessions. The sessions, which took place over a period of one month, were carefully standardised according to day, time, and instruction. Lighting, sound and temperature were monitored and kept constant. The same tester performed all measurements within one hour on the same day each week on both right and left sides of the subjects.

	N (no.)	**MIN**	**MAX**	**MEAN**	**STD.DEV**
HEIGHT (cm)	**5**	**158**	**184.5**	**172.3**	**12.00**
WEIGHT (kg)	**5**	**68**	**100**	**82.4**	**14.24**
AGE	**5**	**31**	**45**	**36**	**5.38**

Table 3.1 Characteristics of subjects

3.2.1. Instrumentation

To conduct this study, two quantitative clinical measuring tools were used- the universal goniometer and a tape measure. Visual observation was not used due to the lack of reliability (Norkin & White 1985; Watkins et al 1991; Youdas et al (1991). The UG was a standard, double-armed goniometer constructed of clear flexible plastic with a scale of one-degree increments. The tape measure was a 100 cm length standard plastic tape measure. In order to reduce the possibility of the Rosenthal effect, which is to prevent the tester from being biased and influenced by their previous readings, tape covered the scales of the goniometer on the side facing the tester and the side of the tape measure facing them was also covered. This allowed an independent recorder to read the reverse side of both instruments and record the measurements. As the goniometer only had measurements in whole degrees, the measurements were rounded off to the nearest degree. The same principle applied to the tape measure in that the measurements for it were rounded off to the nearest millimetre. Using the rounding off effect automatically results in measurement error (Kuzma 1992) but in order to recognize these instruments as reliable, the tester must accept a possible error margin in their measurement (Hogan & Kidd 2001).

There have been many discussions reporting the issues of using both legs in the analysis of gait. For example, if 30 individuals are taking part in a study, (n=30), but if both right and left feet are being used, (n=60). This raises a significant problem according to Menz (2004). He states that it has the potential to significantly influence the findings and interpretation of many investigations. Altman and Bland (1997) also agree that if both right and left feet were counted as single independent observations, the researcher is essentially "double-dipping" their data which essentially means counting each subject twice. Menz (2004) points out that by doing this the "independence assumption of statistical analysis has been violated". Altman and Bland (1997) agree in that if the sample size is inflated, this may lead to spurious statistical significance. Redmond (2004) agrees that although pooling of right and left data cannot be considered best practice, it has to be recognised that one of the downsides of maintaining independence of limbs is that more data will be reported from smaller underpowered studies with the consequence of Type II errors. This could have the potential to result in overly conservative clinical interpretation of results.

Menz (2004) conducted a study using "dummy data" in order to demonstrate how the decision to pool or not pool right and left data can alter results. He developed a dataset of 30 subjects which represented rearfoot motion with and without orthoses in right and left feet. Statistical tests were used to compare the "without orthoses" and "with orthoses" with right and left individually, the average of right and left and right and left combined (n=60). The results of this worked example showed clearly that there are problems in "pooling limbs". There was no effect in rearfoot motion when analysing right foot only, left foot only and when the two were averaged whilst wearing orthoses. However, when the right and left data were combined (n=60), a significant reduction in rearfoot motion was apparent ($p=0.018$, $p < 0.05$). He concluded that depending on whether right and left data were pooled or not, it could be interpretated that foot orthoses either did influence rearfoot motion or did not. It must be remembered that this was only imaginary data but it did provide us with possible evidence of the potential problems with analysing two legs rather than one person.

The only solution around this problem is to select either the right or left side randomly or pick the dominant side or collapse the right and left data into a single measure by taking the average of the two. There is however, a lot of debate on the common practice of deriving data from one leg only (Messier et al 1988; Woodland & Francis 1992; Horton & Hall (1989). Livingstone & Mandigo (1999) report that the measuring and reporting of one value only is problematic for it implies that measurements are bilaterally symmetric. Hahn & Foldspang (1997) conducted a study of 339 athletes and found right Q-angles to be on average 3 degrees greater in magnitude than and significantly different ($p<0.001$) from those in the left lower limb.

For the purposes of the main research, data will not be pooled and each subject will be considered a "unit" of investigation. In doing so, it was accepted that by this approach may result in missing something imperative either by discarding important data or obscuring potentially significant information by taking an average of my measurements. However, it was felt that if one subject is the "unit" of investigation, then it should be the "unit" of analysis (Altman and Bland 1997). However, in this chapter, when assessing intratester reliability and validity of the clinical assessment, it was decided to pool the data from both feet and legs (n=5 subjects which leads to 10 legs, therefore n=10). The statistical justification we have for this reasoning is that we were comparing each variable separately over a period of three occasions. Comparisons were not being sought between subjects or even between right and left limbs therefore using n=10 on this occasion will add strength to the study.

From the table shown below, the ICC results for the average of right and left measurements are much more favourable compared to pairing the two together. This is an example using my data of how this issue has the potential to significantly interfere with results and findings.

MEASUREMENT	**ICC N=10**	**ICC N=5**
ANKLE D/F WITH KNEE EXTENDED	0.72	0.85
ANKLE D/F WITH KNEE FLEXED	0.84	0.88
ANKLE P/L WITH KNEE EXTENDED	0.37	0.44
ANKLE P/F WITH KNEE FLEXED	0.82	0.91
1ST MTPJ DORSIFLEXION	0.91	0.98
KNEE FLEXION	0.91	0.93
KNEE EXTENSION	0.72	0.79
STJN	0.90	0.90
RCSP	0.88	0.97
LEG LENGTH	0.89	0.91
ANKLE WIDTH	0.88	0.90
KNEE WIDTH	0.85	0.90
Q-ANGLE SUPINE	0.70	0.70
Q-ANGLE WEIGHTBEARING	0.70	0.89
REARFOOT ANGLE	0.75	0.94

Table 3.2 Two feet or one person[10]

[10] Redmond 2004

3.2.2. Procedure

Due to the nature of foot abnormalities and the variation within "the normal foot", it is necessary to develop assessment methods that can be consistently applied to all feet and are meaningful and valid biomechanically and clinically. The former requires a measurement and marker placement protocol that can be undertaken on almost all feet and the latter is complicated as axes for inter-segmental motion do not conform to the standard anatomical planes (Inman 1976).

An assessment protocol was devised based on work by Norkin & White (1985). This protocol was used to refine the intratester reliability study prior to data collection. It was postulated that if good to high reliability was proven then the measurement protocol would be used for the main data collection. During each session, subjects were asked to wear shorts therefore their lower extremities were exposed from the level of mid-thigh to their feet. The fifteen measurements are described in detail below.

Measurements to be included in examination

Refer to chapter 2 section 2.5 - Clinical assessment for a full and concise explanation of measurements involved.

1. The Q-angle (supine and weight bearing)

Various authors have stated that a lack of standardised measurement protocol is partially to blame for the doubtfulness of the Q-angles diagnostic value (Woodall & Welsh 1990; Woodland & Francis 1992 and Ando et al 1993). However, this study will follow a strict measuring protocol with each subject being measured in the same foot and knee position in supine and lying positions. This will establish the reliability required in order to be included in the clinical examination. The Q-angle is measured by using a goniometer with the subject in a supine position and a line is drawn from the ASIS to the middle of the patella. Another line is drawn from the middle of the patella to the tibial tuberosity. The Q-angle is the angle between the two lines. It should be measured and compared bilaterally. The same test should be repeated with the subject in a standing position. In the standing position, the subject is asked to find a comfortable resting stance position with their body weight distributed on both feet. It should be emphasised that each subject should stand in a relaxed position and should try to avoid contracture of the quadriceps, which has been proven to decrease the standing Q-angle (Guerra et al 1994; Lathinghouse & Trimble 2000). The Q-angle is marked in the same manner as above. Both positions should then be measured bilaterally and compared with each other.

Figure 3.1 Q-angle performed supine[11]

Figure 3.2 Q-angle performed standing[12]

2. Ankle dorsiflexion/plantarflexion (knee flexed and extended)

As mentioned in section 2.5.2, clinical measurement of ankle ROM may be reliable if a standardised protocol is followed. The same tester was used each time, the occasions were conducted in a short space of time and the same goniometer was used on each occasion.

The subject was positioned prone with their feet hanging over the end of the plinth. The researcher holds the STJ in neutral position as previously described. The position of the STJ and its effect on ankle dorsiflexion is very important. This is because maximum ankle dorsiflexion occurs when the STJ is in its neutral position therefore therapists must measure ankle dorsiflexion with the STJ in its

[11] Smith et al 2008
[12] Smith et al 2008

neutral position (Root et al 1977). If this does not occur and the STJ is in a pronated position, dorsiflexion can occur around the oblique axis of the MTJ that may be confusing and prevent a false measurement of ankle joint dorsiflexion. The knee is extended and the axis of the UG was positioned over the lateral aspect of the centre of the right lateral malleolus. The stationary arm is positioned parallel to the lateral border of the fibula, and the moveable arm was parallel with the lateral border of the fifth metatarsal. The subject then moved the foot into dorsiflexion as far as possible and the recorder measured the angle. The same test was also done with the knee flexed. If dorsiflexion was greater with the knee flexed at 90 degrees than that with the knee extended, gastrocnemious muscle is likely to be contracted and if there was no change with this manoeuvre, there was likely to be a contracture of soleus. This was repeated with the foot moved into plantarflexion. The testing position is the same for ankle dorsiflexion (Rome & Cowieson 1996). The tester gently pushed downward on the dorsum of the subject's foot to produce plantarflexion. The therapist exerted no force on the subject's toes and was careful to avoid pushing the ankle into inversion or eversion. The moveable arm of the goniometer was then aligned and the angle recorded. The same test was also done with the knee flexed. It must be noted that for the purposes of this study, ankle joint range of motion was measured in a non-weight- bearing position.

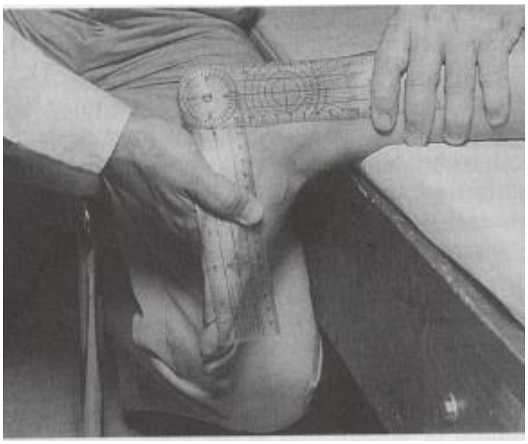


Figure 3.3 Measurement of the ankle dorsiflexion with knee extended[13]

[13] Donatelli 1996

Figure 3.4 Measurement of the ankle dorsiflexion with knee flexed[14]

3. Limb length discrepancy

As discussed on section 2.5.3, it is generally agreed that the tape meansure is the most common measurement tool used clinically to measure LLD despite its low validity and reliability (Freiberg et al 1988). However, due to cost implications and the adverse effects of radiation it still remains the tool of choice. In order to obtain the most reliable results possible using the tape measure, measurements were conducted twice and the mean calculated. Beattie et al (1990) stated that this was the best way to obtain valid and reliable results.

The tester started the measurement from a proximal position at the level of the hips and the subject was in a supine position. The tester positioned the subject's lower limbs in neutral hip rotation as determined by observation then palpated the subject's inferior border of the ASIS which represented a relatively reliable bony landmark of the pelvis. One end of a measuring tape was placed on the ASIS and stretched to the prominence of the medial malleolus on the same side. A measurement was then recorded and compared with the measurement on the contra-lateral side. As with all measurements, it was repeated and the mean calculated. LLD was calculated by subtracting the right leg-length measurement from the left leg-length measurement (Beattie et al 1990; Messier et al 1991).

[14] Donatelli 1996

4. Subtalar Joint Neutral (STJN)

Elveru et al (1988) proposed a method which appears to have construct validity, has shown to have moderately good intratester reliability and appears to be clinically useful therefore this was the method employed for this measurement. The subject is in a prone or seated position with the foot and ankle to be measured hanging over the edge of the couch. The therapist palpates the medial and lateral borders of the lower one third of the leg. With a straight edge, the therapist draws a narrow vertical line on the posterior lower leg. The line should be midway between the medial and lateral borders. The therapist then locates the medial and lateral heads of the talus on the dorsum of the foot. In order to achieve this correctly, the following was accepted. The medial aspect of the talar head is slightly inferior and anterior to the medial malleolus and proximal to the navicular and the lateral aspect of the head of the talus is anterior to the lateral malleolus towards the midline of the foot. To set the foot into STJN position, the therapist places their thumb of the medial hand under the medial aspect of the medial head of the talus and their index finger over the lateral aspect of the talar head. The thumb of the therapists other hand is placed under the plantar surface of the fourth and fifth metatarsal heads. The foot is moved into pronation and supination and whilst the foot is pronating, the talar head can be felt protruding on the medial side of the talonavicular joint and whilst the foot is supinating, the talar head can be felt protruding on the lateral side of the talonavicular joint. Neutral position was achieved when the talar head cannot be palpated on either side (Merriman & Tollafield 1995 and Elveru et al 1988). In this position, the STJ is maximally congruent (Bailey et al 1984). The STJN position was maintained and a UG was used to measure the angle formed by the longitudinal midline of the posterior calcaneus and the midline of the lower leg.

Figure 3.5 Lines showing bisection of posterior heel and lower one third of posterior leg

5. Resting Calcaneal Stance Position (RCSP)

This measurement is clinically important as discussed in section 2.5.6 (Cornwall & McPoil 2004). The RCPS was measured by using the same line drawn when measuring the STJN and the angle this line makes with the ground was measured. This method has been shown to have good intratester reliability and good contruct validity (Elveru et al 1988). With the patient in relaxed standing, the goniometer arms were aligned with the bisection line. The amount of calcaneal eversion or inversion was measured. Values of 0-4°eversion indicated within normal limits, 4-7° eversion indicated the presence of moderate abnormal pronation and eversion greater than 8° indicated marked abnormal pronation (Merriman & Turner 2002).

6. Knee Flexion and Extension

When deciding upon a protocol for knee joint measurement, all favourable and poor reliability results in previous studies were taken into account. As mentioned on p49 of this study, Brosseau et al (2001) found good reliability when measuring knee flexion and extension when using a UG. They recommended using the same tester to take the measurements and the same UG. Many previous authors concur with this theory (Boone et al 1978; Grohmann 1983; Low 1976 and Rothstein et al 1983). Therefore it was deemed very important that a standardised procedure was adhered to using the same tester and the same UG throughout all the measurements. The knee joint measurements were obtained according to the procedure cited by Norkin & White (1985). It was acknowledged that this procedure was not entirely reliable but known sources or error were tried to be reduced thereby increasing the chances of a higher relibability result.

Knee flexion was measured with the subject positioned prone with their ankles and feet hanging off the plinth. This position is crucial to the movement due to the rectus femoris muscle. Measurements were taken in the prone and sitting position. Subjects were asked to actively flex and extend both knees five times before measurement as a warm-up exercise. A stationary arm of the goniometer was placed along the line from the greater trochanter to the knee joint. The moveable arm was placed along the lateral aspect of the fibula (fibular head to lateral malleolus). The axis of the goniometer was located at the intersection of the thigh and shank at the knee joint. They were asked to flex the knee in a posterior direction to full flexion position. The starting and end positions and range of knee flexion were documented. Knee extension was measured with the subject sitting at the end of the plinth with their back straight. The subject started with the knee flexed at 90 degrees and was asked to extend their knee to full extension without altering the position of the pelvis of the lumbar spine. Again, the starting and end position and the range of knee extension were documented.

7. Rearfoot angle-varus/valgus

Literature on the rearfoot angle and its relationship to injury is sparse. Normal values vary as do opinions on the usefulness of the measurement. The justification of including it in the study was to try to find out if there was any relationship between rearfoot angle and any joints proximal to it. It was also included so the angle measured clinically could be compared against the angle taken on radiographs which were conducted later in the study.

Rearfoot angle was measured in relaxed single-limb stance as the angle between a line that bisected the calcaneus and a line which bisected the lower third of the leg. The subject was in a weight- bearing position. The measurement is made under the assumption that in equal, relaxed standing, the subtalar joint should be at or near its neutral position and indicates to the researcher what abnormal compensating may be occurring (Donatelli 1996). A goniometer was placed with its central point at the level of the STJ. One arm of the tractograph was placed on the superior line of the leg and the other arm was placed at the heel bisection line and a measurement was read. The measurement was recorded as degrees of valgus or varus.

8. 1st Metatarsalphalangeal joint (MTPJ) dorsiflexion

The method described below was found to be reliable and valid for use in clinical practice (ICC= 0.87 left foot and ICC=0.97 right foot) using a reliability coefficient by Hogan and Kidd (2001). The midpoint of the medial aspect of the 1st MTP joint was identified by palpation of the joint when the subject was in their normal angle and base of gait. McIlroy and Maki (1997) state that by adopting a preferred stance position, this will meet the need to standardise and the need to simulate 'natural' stance positions during measurement. This technique is recommended by Tranberg & Karlsson (1998) as this reduces the error of skin displacement from a mark made whilst the subject is in a supine or sitting position then used as a reference when they are weight- bearing in stance. Subjects were asked to dorsiflex their hallux to the end range of motion before any examination several times in order to reduce measurement error caused by inflexibility of the joint. The centre of the hinge of the goniometer was aligned with the mark of the medial aspect of the MTP joint, and the proximal arm of the instrument was aligned with the floor. The subject was then asked to dorsiflex their toe to the end ROM. The distal arm of the goniometer was then placed on the dorsal surface of the metatarsal and proximal phalanx and a measurement was read. The same procedure was repeated on the other foot.

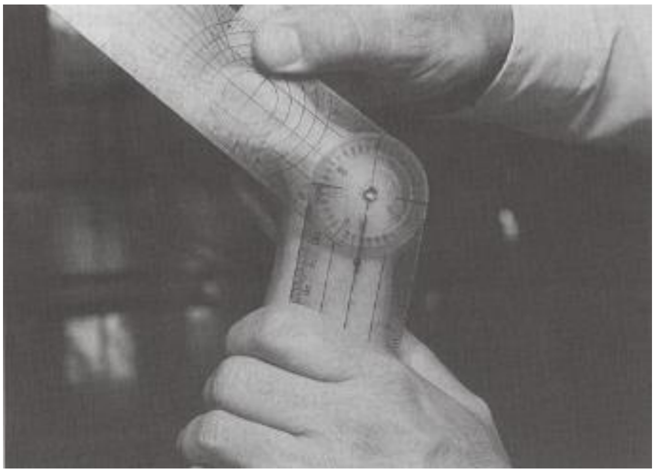


Figure 3.6 Dorsiflexion of the 1st MTPJ[15]

9. Knee width

This was conducted with the subject lying supine with both knees flexed. Knee width was determined from measuring the distance between the lateral and medial femoral condyles in a supine position using a knee alignment device (KAD) of both knees.

10. Ankle Width

This was conducted with the subject lying supine with ankles relaxed and hanging off the edge of the plinth. Ankle width was determined from measuring the transmalleolar distance using a KAD of both ankles.

11. Navicular height/drop

Navicular height and drop was measured using a similar method used by Menz et al (1998) and Evans et al (2003). The tester palpated the medial midfoot in order to locate the navicular tuberosity and this area was marked with a black pen. The fully weight-bearing subject was then placed in the STJN position and asked to maintain this position. The height of the black mark to the supporting surface

[15] Donatelli 1996

was measured with a ruler and recorded in millimetres (mm). Subjects were then asked to relax the foot and the height of the mark to the supporting surface was recorded again. Navicular drop was calculated by subtracting the two measurements from each other and measurements were performed bilaterally.

To ensure the best reliability, the same tester carried out the measurements each time and that said tester had experience in carrying out this measurement procedure. This method allowed for comparison of results with the results from the radiographs conducted later in the study. Radiographs are found to be the gold standard method of measuring LLD (Nichols & Bailey1955).

3.2.3. Statistical analysis

A one-way repeated measures ANOVA was conducted to compare measurement scores over three occasions on the same group of subjects. The reproducibility of all the measurements were considered in this pilot study using the coefficient of variation (CV) expressed as a percentage. This statistic is used to measure overall repeatability. The standard error of measurement (SEM) was used to determine an estimate of the error associated with all repeated measurements over three occasions. Intraclass correlation coefficients (ICC's) were computed to determine the intratester reliability of the researcher. Descriptive statistics were also tabulated.

The data was screened for outliers using box plots in order to determine whether assumptions for repeated-measures ANOVA were met. A probability level of significance of ($p<0.05$) was selected for all statistical tests.

3.3. Results

The intratester variation as a result of the biological variation and the variation of the tester shows for all 15 measurements a mean coefficient of variation of (4.3%). Shown below are sample graphs of mean values of selected variables.

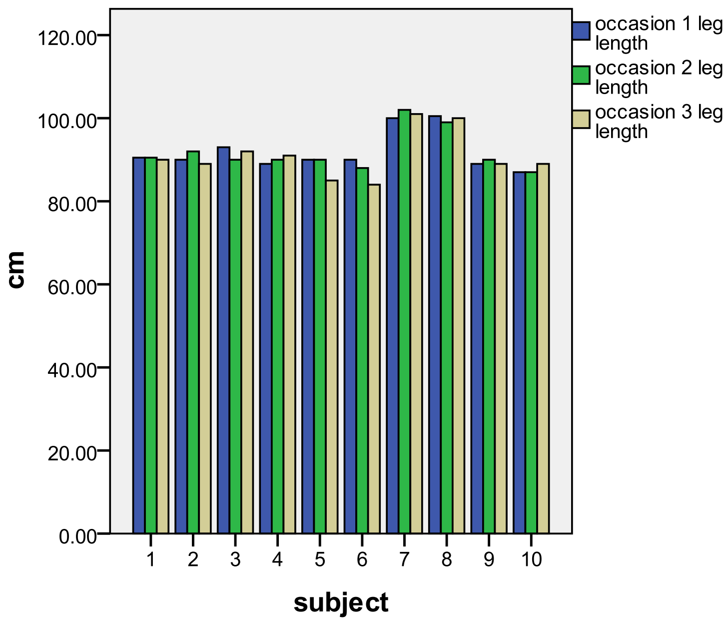


Figure 3.7 Leg length measurements

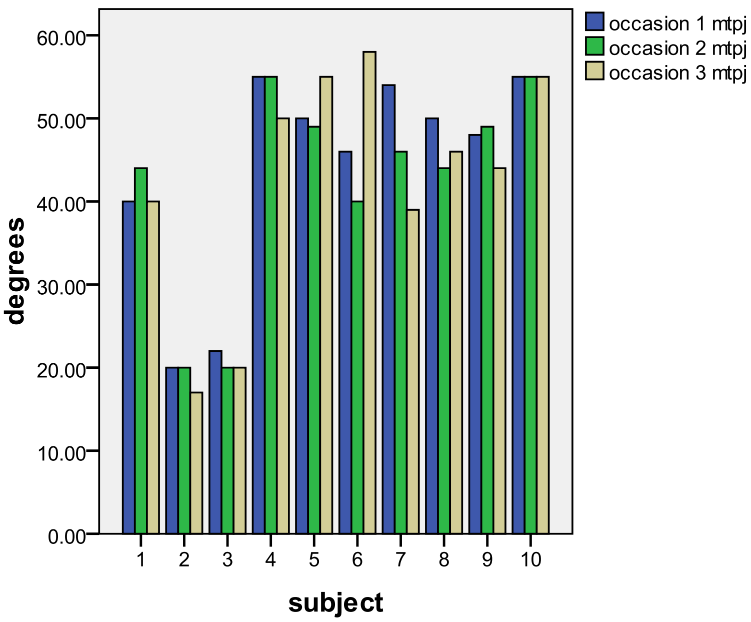


Figure 3.8 1st MTPJ dorsiflexion

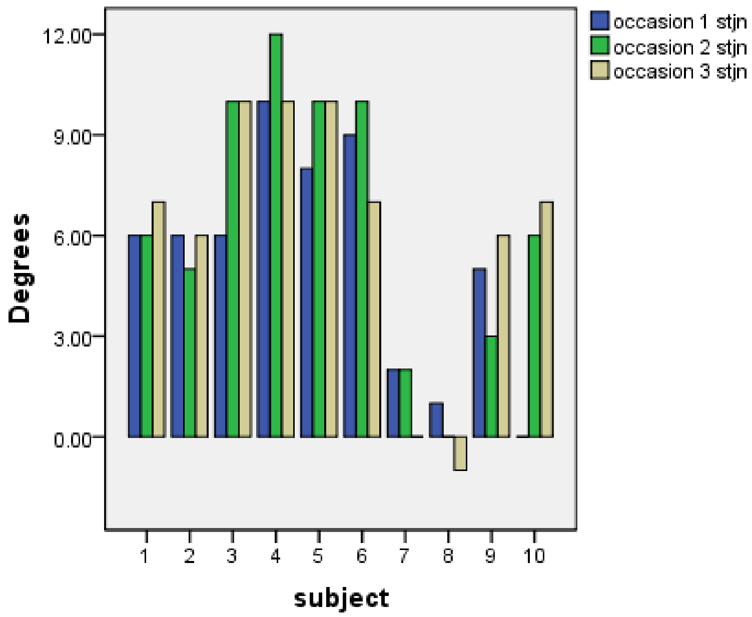


Figure 3.9 STJN measurement

Leg length, STJN and 1st MTP joint measurements all showed very good reliability over the three occasions and from the table below, the CV value is higher for the STJN variable than the leg length variable.

Table 3.3 includes the ICC results, SEM results and CV of all measurements repeated to establish intratester reliability.

MEASUREMENT	**ICC**	**SEM (degrees)**	**CV (%)**
ANKLE D/F WITH KNEE EXTENDED	0.72	0.92	12.06
ANKLE D/F WITH KNEE FLEXED	0.84	1.26	13.47
ANKLE P/F WITH KNEE EXTENDED	0.36	1.91	6.75
ANKLE P/F WITH KNEE FLEXED	0.81	1.35	4.64
1^{ST} MTPJ DORSIFLEXION	0.91	2.07	9.27
KNEE FLEXION	0.91	1.85	3.05
KNEE EXTENSION	0.72	1.16	1.14
STJN	0.90	0.43	26.26
RCSP	0.88	0.46	10.64
LEG LENGTH	0.89	0.98	1.29
ANKLE WIDTH	0.88	0.33	2.2
KNEE WIDTH	0.85	0.17	2.63
Q-ANGLE SUPINE	0.70	0.70	9.3
Q-ANGLE WEIGHTBEARING	0.70	0.68	8.41
REARFOOT ANGLE	0.75	0.63	14.01

Table 3.3 Intraclass correlation coefficients (ICC), Standard error of measurement (SEM) and coefficient of variation (CV) for 15 measurements over three occasions (n=10)

ICC values were found to be significant ($p<0.05$)

3.4. Discussion

The results obtained with the goniometer compare favourably to data in other publications. Brosseau et al (2001) found values of (ICC =0.98) in knee flexion and (ICC = 0.97) in extension. Clapper & Wolf (1988) found ICC values of (0.95) for hip flexion and ankle plantarflexion, (0.92) for ankle dorsiflexion and (0.95) for knee flexion. A list of other previous literature conducted is shown in the table below which is consistent with this present study's results.

Table 3.4 displays examples of previous authors ICC results so it can be shown that the results from the study are consistent with work conducted previously on a normal asymptomatic group of subjects.

MEASUREMENT	**ICC of study**	**INVESTIGATOR**	**PREVIOUS ICC**
KNEE FLEXION	0.91	Watkins et al (2001) Rothstein et al (1983) Clapper et al (1988)	0.99 0.99 0.95
KNEE EXTENSION	0.72	Watkins et al (2001) Rothstein et al (1983) Clapper et al (1988)	0.98 0.91 0.85
LEG LENGTH	0.89	Beattie et al (1990)	0.80
REARFOOT ANGLE	0.75	Elveru et al (1988) Powers et al (1995)	0.86 0.97
ANKLE DORSIFLEXION –knee extended knee flexed	0.72 0.84	Clapper et al (1988) paper does not disclose method of measurement Weaver et al (2001)	0.92 0.98
ANKLE PLANTARFLEXION Knee extended Knee flexed	0.37 0.82	Clapper et al (1988) paper does not disclose method of measurement	0.96
1ST MTPJ DORSIFLEXION	0.91	Hogan & Kidd (2001)	0.87 left 0.97 right
STJN	0.90	Pierrynowski et al (1996)	0.54 foot care specialist and 0.90 student foot care specialist
Q-ANGLE	0.70	Caylor et al (1993) Greene et al (2001)	0.84-0.90 0.14-0.37

Table 3.4 Previous literature showing ICC values

The only low ICC value found was for ankle plantarflexion with knee extended. All of the other measurements proved good to excellent values. No statistical tests exist to compare the differences between ICC's, so visual observations of the data were used to make this determination. Intraclass correlation coefficients for all measurements may be considered good (0.72 or above) with the exception of ankle plantarflexion with the knee extended (ICC=0.37). Youdas (1993) also found lower intratester values for ankle plantarflexion than dorsiflexion (ICC=0.64) was the lowest value. The result of (ICC=0.37) in the present study does not correlate well with the study conducted by Clapper et al (1988) who revealed an ICC of (0.96). However, the coefficient of variation for ankle plantarflexion was (6.75%) which is very acceptable variation. The relatively small interindividual variation explains low reliability coefficients thus measurements with a low CV may be valid but not always reliable measurements. The opposite can also occur and Bovens et al (1990) state that it is common to find a good ICC value where the standard deviations and the CV's are high. They found fairly good reliability coefficients for dorsiflexion of the ankle (0.63-0.75) but high CV's (50.8-10.5%).

Quite a lot of current literature relies on Intraclass correlation coefficients because they are simple to calculate, easy to understand and occur frequently in literature containing reliability of joint range of motion. However, Weaver et al (2001) states that they are non-intuitive and do not directly translate into clinically meaningful estimates of reliability. Weaver et al (2001) and Vinicombe et al (2001) both present ICC's to make their results more comparable to previous literature.

The CV of STJN range is large (0%-100%), not as a result of the standard deviation but because the range of motion is small. Bovens et al (1990) agree that intratester variation strongly depends on the size of the measured angle.

The ANOVA in this pilot study revealed no statistical significant intratester variance (F (27, 2), 20.54, $p=0.37$). This finding supports claims in the literature that an individual tester is capable of making accurate repeated measurements in order to provide reliable information of joint range of motion and function (Low 1976; Boone et al 1978; Clapper & Wolf 1988).

The consistency of the intratester measurements is good. It must be remembered that there will unavoidably be some small changes in the voluntary measurements such as ankle joint ROM, knee joint ROM and 1st MTP joint dorsiflexion from day to day as it is improbable that the subject could achieve exactly the same angle on each of the three occasions. Some part of the variations in the measured angles would be accounted for by in these changes.

3.5. Conclusion

This small study demonstrated that a relatively high degree of intratester reliability can be expected when using the standardised protocol as described above. This fulfils the objective of the analyses of intratester reliability of a detailed clinical assessment and sets a precedent for the reliability of any clinical measurements conducted in the following chapters.

Chapter 4 – Vicon 370 Kinematic Motion Analysis System

4.1. Introduction

Three-dimensional (3D) motion analysis has become a widely used tool for both research and clinical assessments. However, for motion analysis to be valuable, reliable data is essential (Ferber et al 2002). In order to produce reliable and repeatable results, there is a need to minimise the problems associated with the re-application of markers on different test sessions. This is why it was essential to measure the tester's performance in marker application. This ensures that the confidence with which any kinematic 3D results can be used is established (Keenan & Bach 1996).

The main objective of this chapter was to conduct a small pilot study in order to determine the test-re-test and intratester reliability of marker placement using Vicon 370 Kinematic Motion Analysis System and the degree of normal variation during the walking cycle.

4.2. Methodology

Gait analysis was performed using the Vicon 370 Kinematic Motion Analysis System at The Robert Gordon University with six video based cameras (Motion Analysis Corporation) and a floor-mounted force plate (Kistler Instruments, Switzerland). This system has been shown to be one of the most accurate optical measurement systems and is often considered the Gold standard in dynamic motion analysis by many other researchers (Ehara et al 1997; Kadaba et al 1987; Deschamps et al 2011). See section 2.6 on Review of Vicon 370 Kinematic Motion Analysis System and 2.6.2 on Reliability and Validity of Vicon 370 Kinematic Motion Analysis System.

Figure 4.1 Gait laboratory, Robert Gordon University

The researcher who had previous experience in marker placement using the Vicon 370 Kinematic Gait Analysis System carried out reliability analysis on two subjects (not included in the main study). This was conducted to measure the researcher's ability to reliably place the markers on the various landmarks on the body. This was imperative to gain accurate and reliable results for which we can report confidently. The two subjects, who volunteered, were chosen at random. (Subject 1, female, age 45, weight 65kg, height 155cm; subject 2, male age 28, weight 85kg, height 185cm). Both volunteers were free from pain, musculoskeletal and neurological dysfunction of both right and left lower extremities and were familiar with the purpose of the study and signed a consent form prior to starting any sessions.

The researcher placed twenty-six small light reflective markers on various anatomical landmarks three times over a period of three weekly occasions. This was conducted at the same time of day within one hour and attempts were made to keep lighting, sound and temperature constant.

Figure 4.2 Location of markers (Sacrum and posterior heel markers are omitted from the picture)

Once all the markers were applied to each subject, calibration procedures were carried out before any subject data was collected to provide a baseline reference for the 3D coordinate system and to find the best camera orientation (Kejonen & Kauranen 2002). This was carried out using a calibration L- frame initially over the data collection space. The frame contained four retro-reflective markers and was placed in a predetermined position on the motion analysis force platforms. Following this, a dynamic calibration was conducted using a 500mm DYNACAL wand which had two reflective markers on it and involved moving it around the data capture area for approximately twenty seconds.

Calibration may be a main source of error which may reduce the validity of the movements being analysed. The level of kinematic data analysis system calibration can affect the accuracy of the marker placement if it is not configured correctly. The calibration residual scores reflect the cameras ability to record three-dimensional coordinates. A calibration residual score above two can produce inaccurate data coordinates. A reference static trial was then collected of each subject such that the long axis of the foot was aligned in the sagittal plane with the knees in full extension and both arms out to the side. This was done in order to define the neutral position of the joints.

The calibrated measurement field was approximately 5 metres in length by 2 metres in width and was located on top of the force platform.

Each subject was asked to walk barefoot at a natural self-selected speed and was given a few practice trials to determine normal walking pace in order to maintain a consistent average velocity. Subjects were instructed not to target the force plates with their feet as forced kinematic data may result. It was also important for each subject to be familiar and confident with this process as not to alter their gait pattern in any way in order to strike the force plates (Heiderscheidt 2002). A trial was deemed acceptable if both the entire right and left foot of each subject contacted the force platform.

Figure 4.3 Subject walking over force plates

There has been much debate as to the number of trials a subject should perform to reduce the variability that inevitably exists between trials (Diss 2001). Bates et al (1983) stated that a minimum of eight trials was required to establish normal patterns under experimental conditions. However, Grainger et al (1983) and Hughes et al (1991) all disagree and state that at least three gait cycles are all that should be required in order to obtain reliable measurements. Based on work conducted by Diss (2001) and Winter (1984), who state that ten trials are acceptable to obtain reliable results for kinetic data and five trials for kinematic data. They found that subjects demonstrated low variability for kinematic data (CV=7%-20%) but high variability for kinetic data (CV=67%-72%) especially moment force patterns at the knee and hip. Diss (2001) reported that in a study of five normals, kinetic data had a reliability of (ICC = 0.93) when the mean of ten trials were used. This showed that if a subject from the same population performed ten trials, an accurate representation of their kinetics can be gained. They also report that in some cases certain variables and subjects will show inconsistency between trials even when tested under the same experimental conditions despite the number of trials recorded. Devita et al (1988) stated that researchers need to recognise that variability under test conditions could result from "normal performance variability". Therefore conclusions taken from studies with a limited number of trials need to be treated with caution. Winter (1984) suggested that a minimum of three gait trials should be taken to overcome the effects of cycle-to-cycle variability.

Averages of five trials were conducted for each subject for kinematic data. The kinematic data recorded included the pelvis, hip, knee and ankle angles in the transverse, sagittal and frontal planes. Temporal spatial data was also recorded. Kinetic data was not recorded.

Each trial was normalised to 100% (heel strike-heel strike) of the gait cycle to allow for comparison of subjects. Videotaped trials were made simultaneously from the frontal and sagittal planes.

Skin movement was not quantified in this pilot study. However, any errors due to skin movement would presumably be equally visible in all subjects and the results would not introduce any bias. The results, however, must be interpreted with this in mind.

4.3. Marker placement protocol

Palpation was used in this study and was done directly on to the skin over the particular landmark and not over clothes. The skin over the bony landmark was marked using a black felt pen. If the skin above the bony landmark is liable to displacement, marking should be carried out with these bones in the relative position they assume when the anatomical landmark calibration procedure is carried out. For example, ASIS landmarks may be displaced if they are marked when the patient is supine and they are slightly overweight to when they stand up.

Intratester reliability of marker placement is very important to establish especially when reliability is in part dependent on placement of markers by the operator. The following describes in detail where the standard Helen Hayes marker set (Kadaba et al 1987) should be placed on the subject. The positioning is identical for the right side and the left side (see point 2.6.3. on full explanation of the Helen Hayes Marker Set).

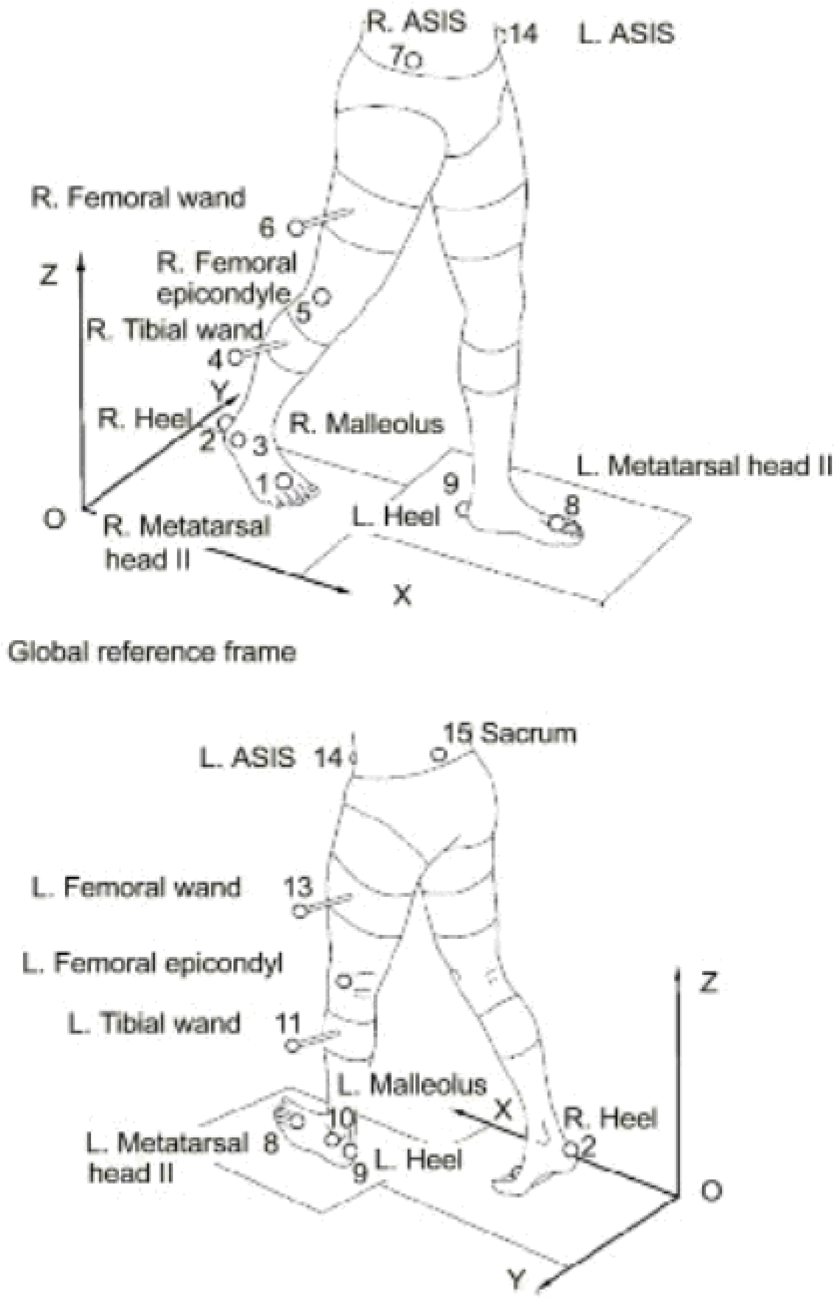


Figure 4.4 Helen Hayes (Davis) Marker Placement Protocol[16]

[16] Lifemodeler 2010

The following anatomical landmarks were reconstructed.

4.3.1. Pelvis

This is defined by three markers

LASI (14)	**Left ASIS**	**Placed directly over the left anterior superior iliac spine**
RASI (7)	**Right ASIS**	**Placed directly over the right anterior superior iliac spine**

Standing in front of the subject, the researcher's hands are placed above the subjects hips. The thumbs are anchored upon the anterior portion of the iliac crest and then follow the downward curve of the iliac crest anteriorly until the anterior superior spines are located. The spines are subcutaneous and relatively easy to palpate however they can be difficult to find in obese subjects.

SACR (15)	**Sacral wand marker**	**Placed on skin mid-way between the posterior superior iliac spines (PSIS)**

The SACR was used as an alternative to the LPSI (left posterior superior iliac spine) and the RPSI (right posterior iliac spine) markers.

4.3.2. Leg Markers

These are defined by two markers, one placed on the thigh and the other over the knee joint. There is a third 'virtual 'marker which is the calculated hip joint centre. The three sets of coordinates generate a plane which is perpendicular to the plane of hip flexion/extension.

LKNE (12)	**Left knee**	**Placed on the lateral epicondyle of the left knee**
RKNE (5)	**Right knee**	**Placed on the lateral epicondyle of the right knee**

In order to locate the "precise" point for the knee marker placement, the subject was asked to passively flex and extend their knee slightly. This was done to identify where the knee joint axis passes through the lateral aspect of the knee and to find the lateral skin surface that comes closest to remaining fixed in the thigh. This area was also the point about which the lower leg appears to rotate. The area was marked with a pen and was approximately 1.5cm above the knee joint line, mid-way between the front and the back of the joint. The markers were attached at these points.

LTHI (13)	**Left thigh**	**Placed over the lower lateral 1/3 surface of the left thigh, just below the swing of the hand**
RTHI (6)	**Right thigh**	**Placed over the lower lateral 1/3 surface of the right thigh, just below the swing of the hand**

The anterio-posterior thigh markers are used to calculate the knee flexion axis location and orientation. It is critical that these markers are aligned correctly. The thigh markers contain a 5cm broad base plate as their placement is over soft tissue but it is important to keep the markers off the muscle belly of the thigh. The markers also have a "wand" and this was adjusted so that it was aligned in the plane of the knee and hip joint centers.

LANK(10)	**Left ankle**	**Placed on the left lateral malleolus along an imaginary line which passes through the transmalleolar axis**
RANK (3)	**Right ankle**	**Placed on the right lateral malleolus along an imaginary line which passes through the transmalleolar axis**

The lateral ankle markers were placed over the lateral malleolus along the dorsiflexion/plantarflexion axis of the ankle.

LTIB (11)	**Left tibial wand marker**	**Placed over the lower 1/3 of the left tibia to determine the alignment of the ankle flexion axis**
RTIB (4)	**Right tibial wand marker**	**Placed over the lower 1/3 of the right tibia to determine the alignment of the ankle flexion axis**

The tibial markers are very similar to the thigh markers and were placed on the plane which contained the knee and ankle joint centers and the ankle dorsiflexion and plantarflexion axis i.e. the lateral knee marker and the lateral ankle markers.

4.3.3. Foot Markers

LTOE (8)	**Left toe**	**Placed on the left dorsum of the foot on the 2nd ray**
RTOE (1)	**Right toe**	**Placed on the right dorsum of the foot on the 2nd ray**
LHEE (9)	**Left heel**	**Placed on the left posterior calcaneus**
RHEE (2)	**Right heel**	**Placed on the right posterior calcaneus**

The foot markers were placed on the dorsum of the foot on the 2nd ray at the metatarsal heads and on the posterior calcaneus such that its height was equal to that of the marker on the dorsum of the foot and they were parallel to the sole of the foot in reference to the floor. This was to establish the long axis of the foot (Rogers 1988). Subjects were asked to flex their toes in order to facilitate identification.

Since the motion of the talus cannot be detected in vivo directly with a surface marker set, the markers on the posterior calcaneus is a representation of the talus. This is a valid approach for the normal ankle joint as calcaneal motion can be attributed primarily to subtalar rotation (Simon et al 2006).

4.4. Foot model

4.4.1. Navicular height

Little is known regarding the effects of orthotics on the mid-foot due to the difficulty in measuring these mechanics with surface markers (Stackhouse et al 2004). With the surface marker on the navicular bone, this provided extra information on mid-foot mechanics. To date, Simon et al (2006) is the only study conducted with a skin based navicular marker. There are however limitations of using a navicular marker according to them. The tibialis anterior tendon may affect the correct placement of it and may have to be attached in a more medial or lateral position. Regardless of this possible anatomical variation, the navicular marker provides valuable information needed for mid-foot kinematics.

Figure 4.5 Subject's foot with extra navicular marker

LNAV	**Left navicular**	**Positioned on the most prominent aspect of the left navicular tuberosity.**
RNAV	**Right navicular**	**Positioned on the most prominent aspect of the right navicular tuberosity.**

4.4.2. Rearfoot angle

An estimation of the motion of the calcaneus with respect to the tibia in the frontal plane is often used to produce two-dimensional values for inversion and eversion (Rodgers & LeVeau 1982; Bates et al 1978). Many studies investigating foot function during gait have approximated foot pronation and supination using calcaneal eversion and inversion (Cavanagh 1990). Pronation was defined as the eversion angle between the bisection of the distal 1/3 of the lower leg and the bisection of the calcaneus. Supination was defined as the inversion angle between the bisection of the distal 1/3 of the lower leg and the bisection of the calcaneus.

When using the Helen Hayes marker set, it was discovered that some of the variables of interest could not be measured within this marker set. Inversion and eversion values (pronation and supination) were required to measure the amount of pronation with and without shoes and orthotics. Inversion and eversion are frontal plane motions of the ankle and pronation and supination are triplanar motions of the foot and ankle complex. Pronation is made up of dorsiflexion, abduction and eversion motions and inversion is made up of plantarflexion, adduction and inversion motions. As inversion and eversion are components of supination and pronation and can therefore be used interchangeably with supination, as can eversion with pronation. By placing the extra heel marker on the lower third of the calcaneus along with the original Helen Hayes heel marker, measurement of 2D frontal plane inversion and eversion of the rearfoot was achieved. A mathematical model was also written in Bodybuilder program. Bodybuilder software was designed as a scripting language which includes full support for advanced concepts such as segments, vectors, angles, moments and powers. This allows the user to define a complete segment or to calculate joint moments very simply (Vicon Motion Systems 2011). The model was written which projected a line containing the two posterior calcaneal markers and the two proximal markers on the back of the leg. The 2D rearfoot angle was then calculated as the angle formed by the intersection of these two lines. Measurement of the inclination of the calcaneus relative to the lower leg in resting standing position frequently referred to as the rearfoot angle. Its clinical importance is related to the premise that excessive foot pronation is manifested by an everted calcaneus during quiet standing (Donatelli 1996).

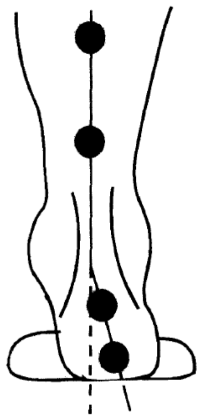


Figure 4.6 2D Rearfoot angle measured during walking in the present study[17]

Figure 4.7 Location of markers on posterior leg and heel

[17] Cornwall & McPoil 1995

4.4.3. 3D Calcaneal Inversion/Eversion angle

Many studies have used 2D motion of inversion and eversion data which have then been used to estimate and present 3D motion of eversion and inversion (Bates et al 1979 and Winter 1984). Because of this it was decided to create a new kinematic model which would produce an actual 3D value. Kinematic modelling is the process of transforming 3D points obtained from reflective markers to biomechanical measurements such as joint angles. By creating segments aligned with anatomical axes, it is possible to generate accurate joint angles either between two segments or relative to the laboratory coordinate system (Vicon Motion Systems 2011). A segment was created of three points, one being the original lateral malleolus, one being the original posterior heel and the third point was made from creating a virtual marker on the medial calcaneus.

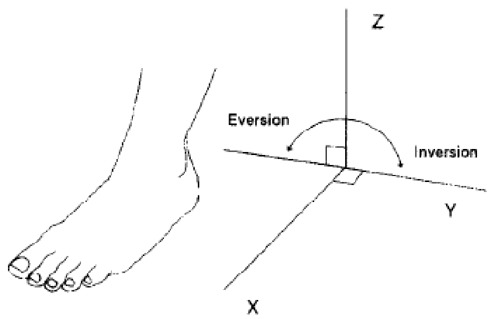


Figure 4.8 3D Eversion-inversion foot angle measured during walking in the present study[18]

Subjects were asked to lie prone on the plinth with their feet hanging over the edge and the calcaneus was visually bisected and a black line was marked. Procedures in how to do this have been discussed in an earlier chapter. The calf was also bisected visually and marked. Lapointe et al (2001) conducted a study which examined the reliability of bisecting the calcaneus using digital linear callipers and the visual bisection technique. They found very disappointing results with visually palpating the bisection to be poor (intrarater ICC= 0.30). The digital callipers provided a mean absolute error of less than 1° which for the purpose of the bisection of the heel, this was considered highly repeatable. However, Elveru

[18] Cornwall & McPoil 1995

et al (1988) chose to visually bisect the calcaneus because they found that a calcaneal bisection line did not accurately represent calcaneal position during subtalar joint range of motion measurements secondary to skin movement when the joint was moved. They state that visually and objectively measuring the bisection is a far more utilised and accurate method. For the purposes of this present study, callipers were not available within the gait laboratory so the method by Elveru et al (1988) was utilised.

LCALC 1	**Left calcaneal marker 1**	**Positioned 2cm below the L heel marker measured centre to centre**
RCALC 1	**Right calcaneal marker 1**	**Positioned 2cm below the R heel marker measured centre to centre**

One extra small marker with 1cm diameter was placed over the calcaneus on the bisection line 2cm below the larger marker measured centre to centre.

LCALC 2	**Left calcaneal marker 2**	**Positioned 4inch above the left proximal calf marker**
RCALC 2	**Right calcaneal marker 2**	**Positioned 4inch above the right proximal calf marker**
LCALC 3	**Left calcaneal marker 3**	**Positioned 4inch above the left distal calf marker**
RCALC 3	**Right calcaneal marker 3**	**Positioned 4inch above the right distal calf marker**

Another two small markers were placed on the calf measuring centre to centre along the bisection line. The distal marker was placed 4 inches above the proximal calcaneal marker and the second calf marker was placed 4 inches above the distal calf marker.

Figure 4.9 Subject showing extra leg and heel markers

The markers defined two lines which were used to describe the frontal plane angle of the calf-calcaneus angle which is normally referred to in the medical literature as the rearfoot angle (McPoil & Cornwall 1994; Liu et al 1997 and Mosley et al 1996). This was described as the maximum value of the calf-to-calcaneal angle in the direction of pronation recorded during the stance phase. This angle however, reflects only certain aspects of pronation namely changes in the rearfoot in the frontal plane. Pronation in gait includes movement in the sagittal and transverse planes at the STJ as well as movements in the frontal, transverse and sagittal planes at the MTJ.

For the trials performed whilst wearing shoes and shoes with orthoses, the calcaneal marker was applied at the top of the heel counter. The height of the heel marker was measured to reflect the alignment of the sole of the foot and then the marker on the dorsum of the shoe was placed at the same height, proximal to the distal toe crease of the shoe, approximately between the 2nd and 3rd metatarsals (Oeffinger et al 1999).

Cornwall & McPoil (1995) compared 2D and 3D measurements of rearfoot inversion and eversion during running. They reported that the patterns between them were very similar and the areas where the curves are significantly different are during the initial 6% of the stance phase. They concluded that a likely possibility for this difference is that the angle of the rearfoot at the initiation of the stance phase is not accurately measured during the 2D videography. However, there were discrepancies in the rearfoot angle at heel-strike between the two conditions and they did not compare the influence of the two analysis methods on rearfoot motion during the last 40% of the stance phase (Pohl 2006). McClay and Manal (1998b) provided supporting evidence that the rearfoot angle at heel-strike was different between 2D and 3D analyses. They also found that the rearfoot angle at toe-off was influenced by the type of analysis, therefore, making 3D analysis essential for any investigation of rearfoot kinematics over the entire stance phase.

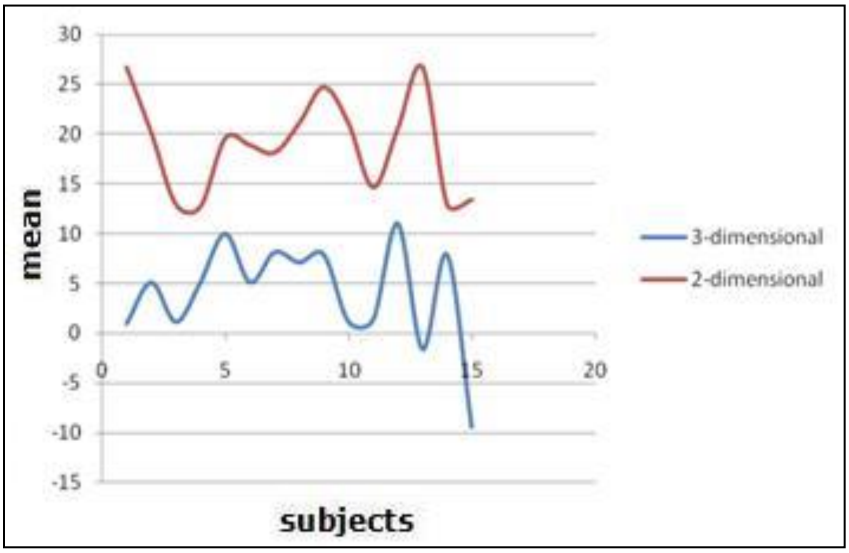


Figure 4.10 2D and 3D rearfoot motion values at heel strike

This graph shows that the the values at heel strike are completely different between the two comparisons ($p<0.05$) which is consistent with McClay and Manal (1998b).

4.5. Statistical analysis

The methodology of data analysis was as used by Kadaba et al (1989) in their well known study on normal adults.

There is no Gold standard method for measuring reliability of marker placement. Many authors have used CMC's (Kadaba et al 1989; Winter 1971; Steinwender et al (2000); Gorton et al 1997; Yavuzer et al 2008; Mackey et al 2005; Kavanagh et al 2005) which compares the waveforms of the kinematic graph. However, for this present study, comparisons of the peaks and troughs of the waves were considered more important and relevant than the pattern of the wave. For this reason the coefficient of variation (CV) was used. The CV is particularly useful for representing the repeatability of performance tests (Yavuzer et al 2008).

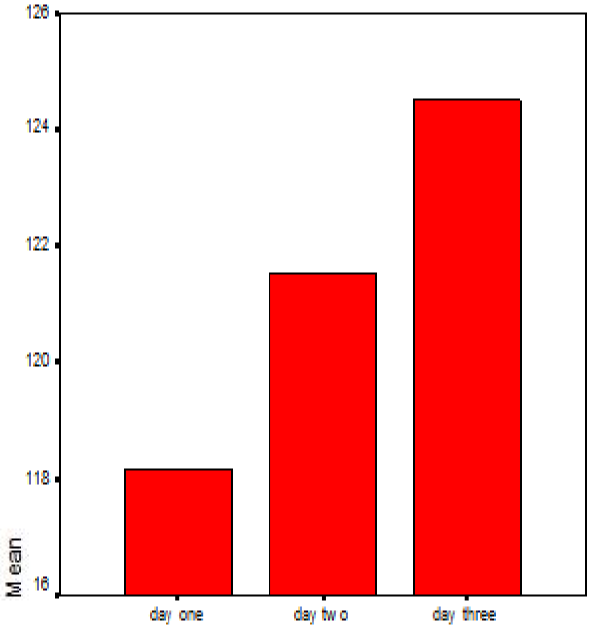
For both subjects, the coefficient of variation (CV) was calculated as a measure of between-day and within-day repeatability. The CV can be defined as the ratio of the standard deviation to the mean value (Steinwender et al 2000). For within-day CV, all gait parameters from all six trials were used, whilst all eighteen trials were used to calculate between-day CV.

Mean and standard deviations (SD) of the spatiotemporal parameters were computed for within each test day as well as over the three test occasions. Visual interpretation of the graphs was facilitated by superimposing the subjects' results onto a normal database (database was from the group of 5 control subjects in chapter 5). The values found at certain points throughout the gait cycle on all three occasions were compared using an independent t-test ($p<0.05$).

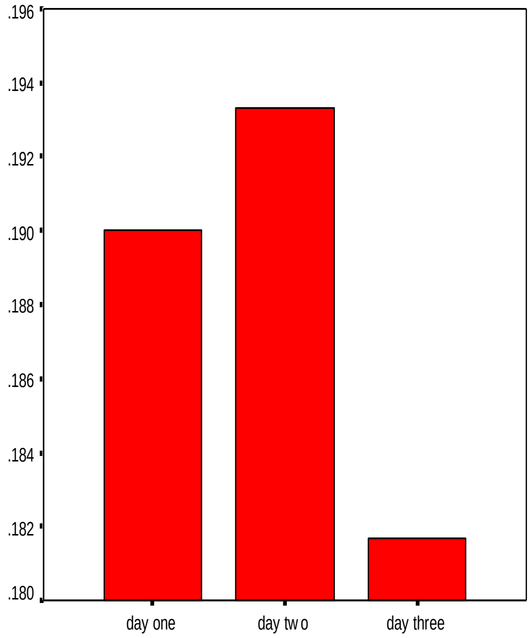
Intraclass correlation coefficients (ICC (2, k)) were used to compare reliability between occasions for all kinematic and spatiotemporal variables. The ICC is a measure of correlation that considers variance and describes the agreement between the repeated measures. This approach is an appropriate statistical method to analyse agreement between sets of data for any sample size (Yavuzer et al 2008). The evaluation criteria and standards for ICC values were as follows, values (> 0.75) indicated excellent reliability, (0.4-0.74) indicated adequate reliability and (< 0.40) indicated poor reliability (Salter et al 2005). Although ICC's are generally considered to be an appropriate indicator of reliability, it must be noted that a low ICC does not necessarily mean that a test is unreliable. In cases where there is little variability among subjects' scores between tests and re-tests, it can be difficult to obtain high ICC's (Menz et al 2003). To overcome this, the coefficient of variation (CV) was also used. This absolute measure of reliability used along with ICC's can show that if a clinical measurement has a moderate or low ICC, it still may be reliable provided that the CV is low and there is no systematic difference between one trial to the next (Menz et al 2003).

4.6. Results

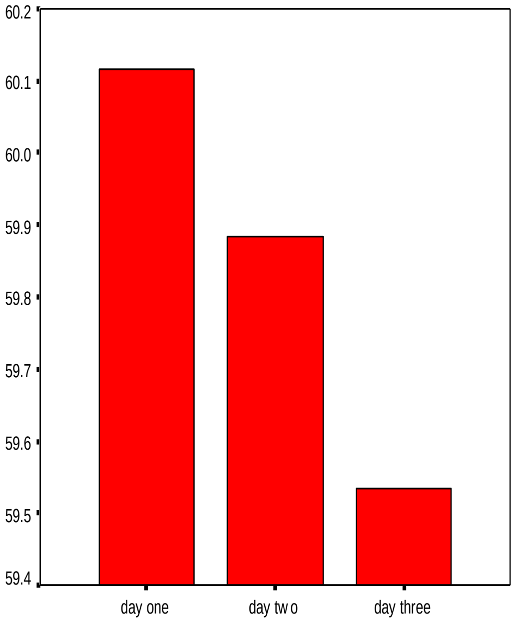
The following results are graphs depicting the mean temporal spatial parameters over three occasions for one subject.



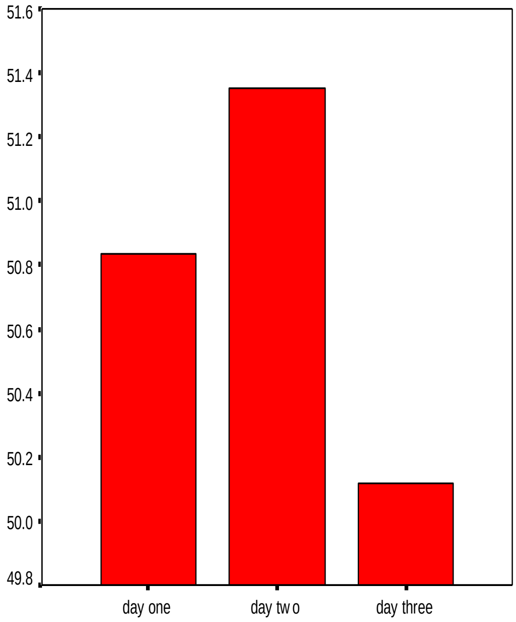
Cadence (steps/min)



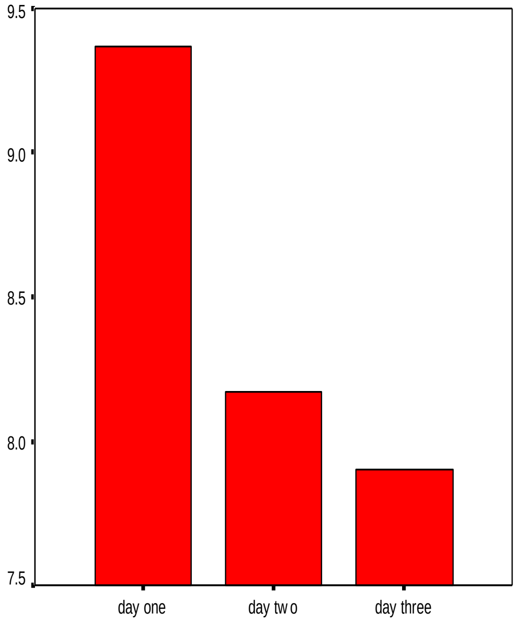
Double support (s)



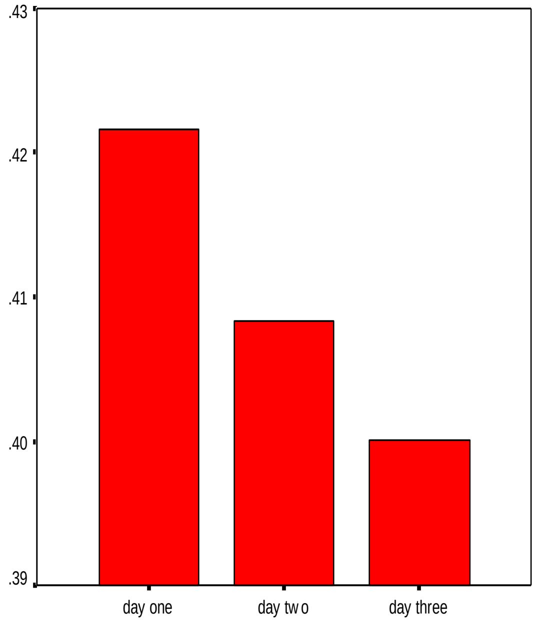
Foot off (%)



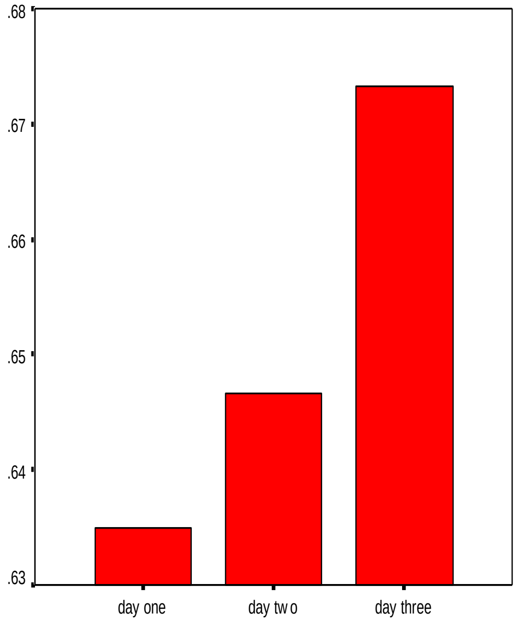
Opposite foot contact (%)



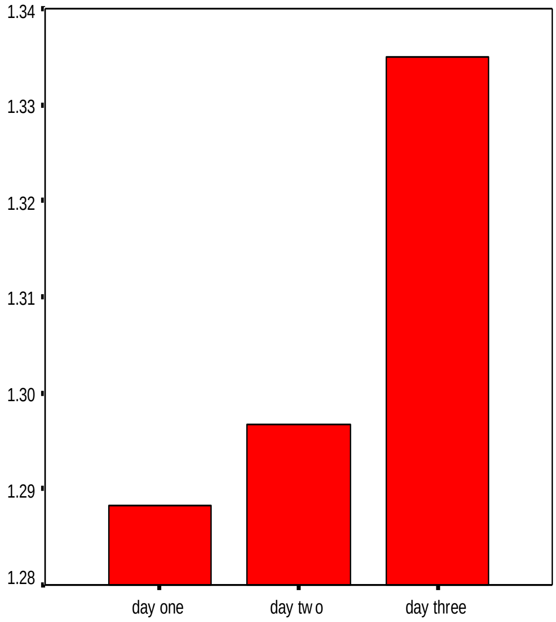
Opposite foot off (%)



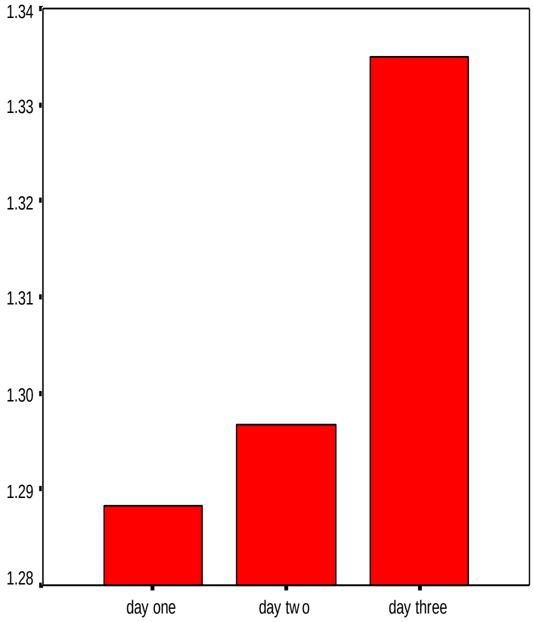
Single support (s)



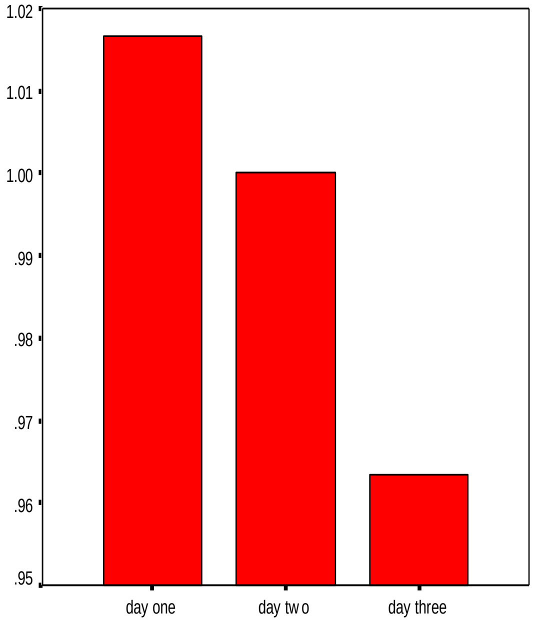
Step length (m)



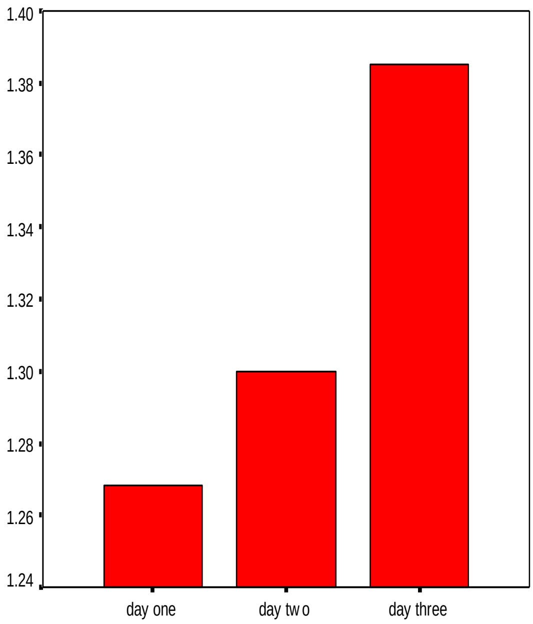
Step time (s)



Stride length (m)



Stride time (s)



Walking speed/velocity (m/sec)

Figure 4.11 Mean of spatio-temporal parameters within each test day as well as between test days for one representative subject

The graphs above show one subject only picked randomly as the results for the other subject were very similar and it must be noted that we are not measuring repeatability between the two subjects but each one separately on different occasions. The graphs depict the fact that the subject walked repeatedly on the three different testing sessions.

The mean and SD of within-day and between-day repeatability of the spatio-temporal parameters reflected by the corresponding CV in the tables below.

			Mean	St Dev	CV %
			Mean	Mean	Mean
Cadence	**Group**	**Within Left**	**121.44**	**3.95**	**3.26**
		Within Right	**121.89**	**2.86**	**2.33**
		Within Days	**121.67**	**3.61**	**2.98**
		Between 3 days	**121.67**	**4.39**	**3.61**
D/ Support	**Group**	**Within Left**	**0.19**	**0.01**	**6.24**
		Within Right	**0.18**	**0.01**	**5.72**
		Within Days	**0.18**	**0.01**	**5.70**
		Between 3 days	**0.18**	**0.01**	**6.09**
Foot Off	**Group**	**Within Left**	**60.43**	**1.4**	**2.29**
		Within Right	**59.02**	**0.87**	**1.48**
		Within Days	**59.73**	**1.47**	**2.45**
		Between 3 days	**59.73**	**1.68**	**2.82**
Opposite Foot Contact	**Group**	**Within Left**	**51.18**	**1.68**	**3.25**
		Within Right	**49.93**	**0.83**	**1.66**
		Within Days	**50.56**	**1.46**	**2.87**
		Between 3 days	**50.56**	**1.63**	**3.23**
Opposite Foot Off	**Group**	**Within Left**	**6.69**	**3.63**	**54.26**
		Within Right	**9.20**	**0.59**	**6.39**
		Within Days	**7.94**	**2.86**	**39.39**
		Between 3 days	**7.94**	**3.25**	**40.95**
Single Support	**Group**	**Within Left**	**0.41**	**0.01**	**2.43**
		Within Right	**0.4**	**0.00**	**1.06**
		Within Days	**0.41**	**0.01**	**2.56**
		Between 3 days	**0.41**	**0.02**	**3.68**

Step Length	**Group**	**Within Left**	**0.66**	**0.02**	**3.73**
		Within Right	**0.65**	**0.01**	**2.00**
		Within Days	**0.65**	**0.03**	**4.72**
		Between 3 days	**0.65**	**0.04**	**5.46**
Step Time	**Group**	**Within Left**	**0.48**	**0.03**	**5.45**
		Within Right	**0.50**	**0.02**	**3.41**
		Within Days	**0.49**	**0.03**	**5.94**
		Between 3 days	**0.49**	**0.03**	**6.46**
Stride Length	**Group**	**Within Left**	**1.31**	**0.03**	**2.38**
		Within Right	**1.31**	**0.03**	**2.14**
		Within Days	**1.31**	**0.03**	**2.51**
		Between 3 days	**1.31**	**0.04**	**3.08**
Stride Time	**Group**	**Within Left**	**0.99**	**0.03**	**3.14**
		Within Right	**0.99**	**0.02**	**2.35**
		Within Days	**0.99**	**0.03**	**2.98**
		Between 3 days	**0.99**	**0.04**	**3.68**
Walking Speed	**Group**	**Within Left**	**1.32**	**0.02**	**1.75**
		Within Right	**1.33**	**0.04**	**2.95**
		Within Days	**1.33**	**0.03**	**2.18**
		Between 3 days	**1.32**	**0.05**	**4.11**

Table 4.1 Means, standard deviations, coefficients of variations for spatio-temporal parameters

The table above clearly shows that within-day is more repeatable than between-day spatio-temporal parameters which strongly agree with Kadaba et al (1989). The coefficient of variation on the left side has a lower CV for all parameters with the exception of walking speed/velocity which has a higher percentage. This was also noted in the study by Kadaba et al (1989).

Among the assessed spatio-temporal parameters, opposite foot off showed the most variability (CV =39.39%) and walking speed/velocity showed the least (CV= 2.18%) for within-day sessions. Between –day sessions showed opposite foot off the most variable again (CV =40.95%) and foot off, stride time and stride length to be the most reliable (CV =2.82; 3.68 and 3.08% respectively). It must be noted that the values for opposite foot off are very high and most unusual and does not conform to other similar studies. It must be assumed therefore that there was a measurement error with this particular variable. The next two most variable parameters were the double support with a CV value of (5.70%) within-day and (6.09%) for between day and the step time with a CV value of (5.94%) within–day and (6.46%) for between days.

Kinematic angles were also measured over the three occasions and samples of graphs are shown below. The peak values of all kinematic angles on all three occasions showed excellent repeatability (ICC= 0.98-0.99), the minimum on all three occasions showed excellent repeatability (ICC =0.78-0.99) with the exception of day 1 which only showed adequate reliability (ICC =0.55). The ICC's show that day 1 was the most variable by a great number. As expected, between-day marker placement ICC values were less reliable than within-day values however they still demonstrated excellent repeatability between the three testing sessions (ICC= 0.79-0.99).

	DAY 1 **Average measure**	**DAY 2** **Average measure**	**DAY 3** **Average measure**
MAX	**0.999**	**0.998**	**0.994**
MIN	**0.786**	**0.998**	**0.974**

Table 4.2 ICC results for within-day marker placement reliability

	AVERAGE MEASURE
MAX	**0.991**
MIN	**0.977**

Table 4.3 ICC results for between-day marker placement reliability

A full list of graphs can be found in the appendix but shown below are a sample of value over three occasions for two subjects.

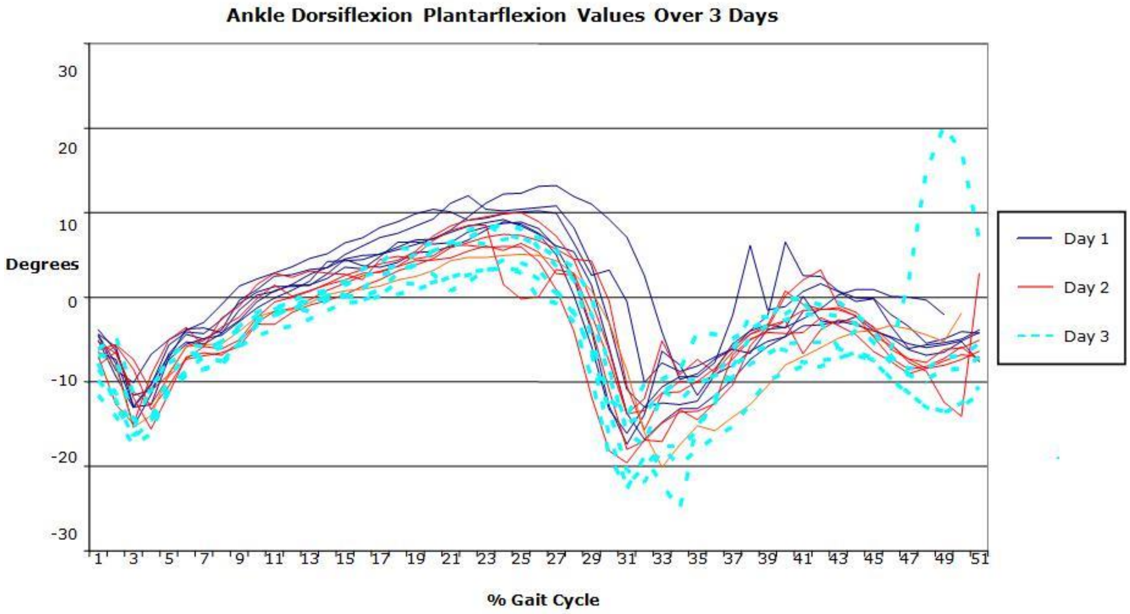


Figure 4.12 Ankle dorsiflexion/plantarflexion values over three days

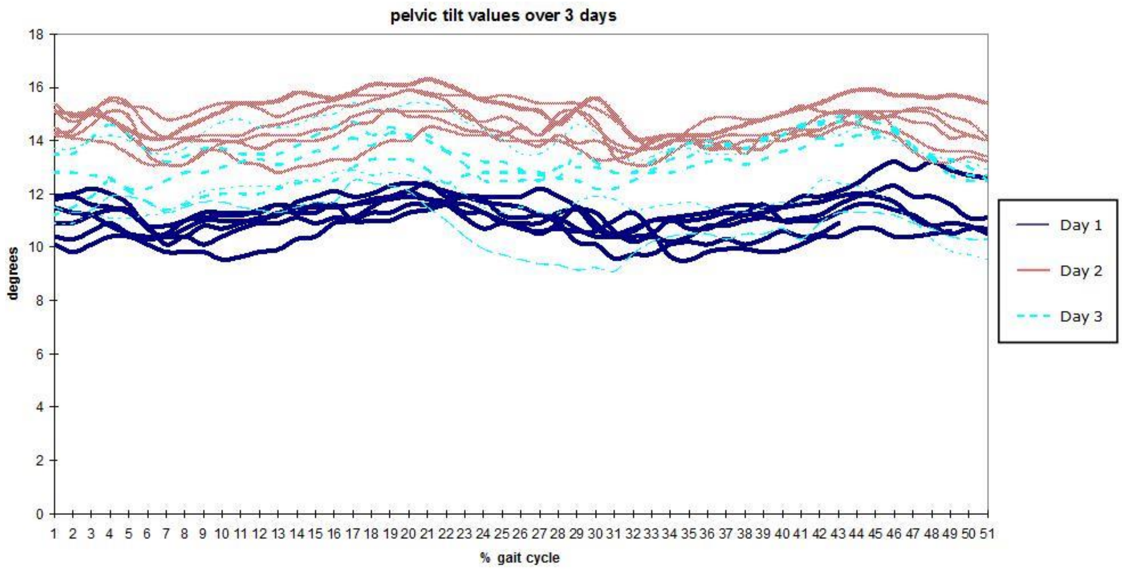


Figure 4.13 Pelvic tilt values over three days

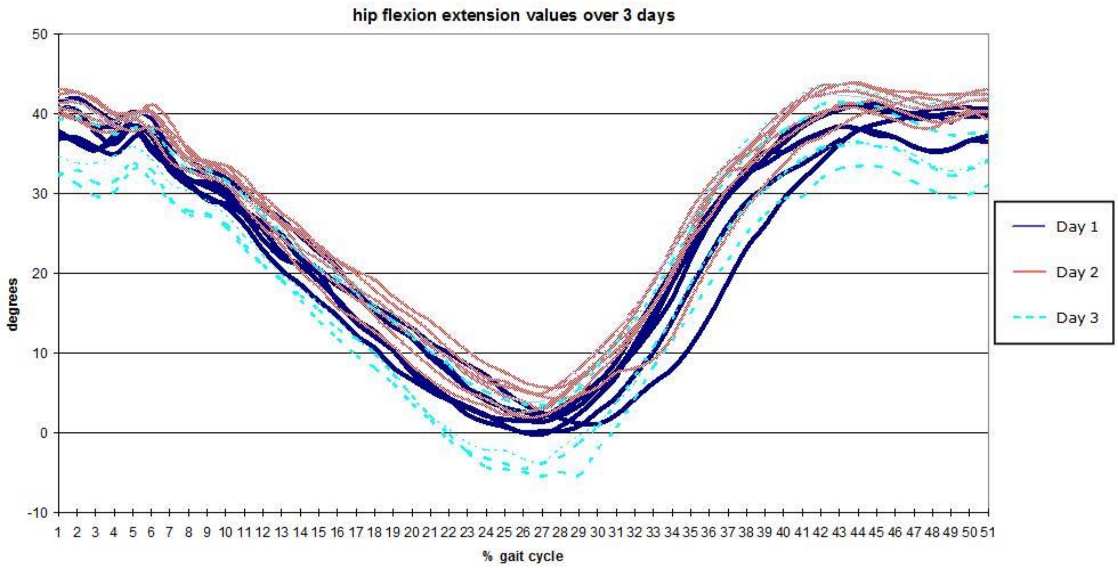


Figure 4.14 Hip flexion/extension values over three days

4.7. Discussion

The purpose of this study was to determine the test-retest and intratester reliability of marker placement of the researcher using the Vicon 370 Kinematic Motion Analysis System. The results of this investigation demonstrate that all the markers were placed in a repeatable and reliable fashion both within and between testing days.

4.7.1. Kinematic parameters

The general patterns of the kinematic parameters of the lower extremity during walking that were observed in this study were consistent with those reported in the literature (Kadaba et al 1989; McFadyen & Winter 1988).

It can be shown from the kinematic graphs that sagittal plane values were less variable than frontal and transverse plane values. Ferber et al (2002) agree with this and found the latter two to be in agreement with each other. They suggest that variability due to marker reapplication is less evident in the sagittal plane. Kadaba et al (1989) also agree with this statement and demonstrated relatively low reliability in frontal and transverse plane values as compared to sagittal plane values in both within and between day comparisons. The variability in any plane motion can be attributed to the alignment of the wand markers and their effect on the eventual kinematic data. Skin movement artefacts are known to exaggerate joint angles particularly measures of transverse plane rotations (Holden et al 1997).

Manal et al (2000) reported in a study comparing various marker configurations to bone anchored markers on the leg that the greatest error between bone and skin attached markers were in the transverse plane. Growney et al (1997) also found that the sagittal plane angles for knee, hip and ankle demonstrated excellent repeatability within test days and between test days compared with fairly repeatable results for frontal and transverse plane angles. Comparisons were not made with the values in this study however as a differing statistic was used to test repeatability.

The factors contributing to between-day repeatability would include the variability in the gait pattern and in the application of markers on the same subject on different test days. To exclude this variability, the markers were applied by the same person on each occasion. In spite of extreme care and a standardised protocol of marker application, some inconsistency is inevitable especially with stick markers for the shank and the thigh and in the placement of the sacral marker. Kadaba et al (1990) and Kadaba et al (1989) attributed errors in the re-application of markers to be a major cause for variations in the gait analysis data. Results of repeatability of kinematic data in our study are in agreement with the conclusions drawn by Kadaba et al (1990).

When comparing the within-day and between day ICC values, it was not surprising that the within-day values were higher than the between day values. This is in agreement with various other studies (Carson et al 2001; Steinwender et al 2000 and Kadaba et al 1989).

The ICC values for maximum and minimum values over the three occasions show that day 1 was the most variable by far. This was also shown by the CV values. Day 1 CV values were much higher than days 2 and 3. Within day variability can be attributed to measurement error, skin marker movement and inherent physiological variability during human locomotion. However, the results suggest that between-day variability can be affected by factors affecting within-day variability such as marker reapplication error (Steinwender et al 2000). Even slight changes in marker placement may produce an offset shift in the data resulting on lower between- day variability.

Although the within-day repeatability of gait data has been shown to be generally good in studies by Kadaba et al (1989) and Growney et al (1997), some variation between trials is to be expected. In some cases, certain variables and subjects will show inconsistency between trials even under exactly the same experimental conditions regardless of the number of trials recorded (Diss 2001). Consequently some variation of the small differences observed in our data may be the result of the natural variation in gait.

4.7.2. Temporal-spatial parameters of gait

Three previous well known studies have reported on the repeatability of the spatio-temporal parameters of gait using the Vicon Kinematic Motion Analysis System. Table 4.4 illustrates the comparisons between them and the present study's results.

Parameter	Kadaba et al (1989) (CV)	Steinwender et al (2000) (CV)	Growney et al (1997) (CV)	Present study (CV)
Cadence (steps/min)	1.9	3.4	2.29	2.98
Walking speed/velocity (m/sec)	2.9	5.2	2.69	2.18
Left foot off (%)	NR	NR	1.80	2.29
Right foot off (%)	NR	NR	1.58	1.48
Left stride length (m)	1.7	4	2.83	2.38
Right stride length (m)	1.7	4	2.83	2.14
Left step length (m)	NR	NR	2.88	3.73
Right step length (m)	NR	NR	2.84	2.00

Table 4.4 Comparison of average CV% values in this study with those of previous studies

KEY: NR = NOT REPORTED

In the above comparisons of previous studies, healthy subjects walked at self-selected normal walking speed. The results from the study by Steinwender et al (2000) are slightly higher than the others including the present study. The reason for this could be that they used healthy children as subjects (aged 7-15) and the others chose adults.

It can be seen that the CV values from the present study compare well with the previous studies and it can be concluded that the healthy subject's gait at a normal self selected walking speed was as repeatable as that of previous reported samples.

It can also be seen that the repeatability of the other parameters not reported in previous studies is reasonable. If the misleading value for left opposite foot off is ignored and the other parameters which are the most concentrated on, double support (CV =5.07%) for within-day and (CV= 6.09%) between-day has high values. This is the shortest event in the gait cycle (Whittle 2002) thus making it susceptible to measurement error (Wall & Crosbie 1996). This just highlight one of the disadvantages mentioned previously of using the coefficient of variation with small mean values- a higher CV value does not necessarily mean low reproducibility (Growney et al 1997).

It must be acknowledged that there may be limitations to the findings of the study because of the use of skin markers. The surface markers were attached with care in order to reduce the effect of skin motion artefact and marker "wobble" errors have not been measured but will be accounted for in this study.

4.8. Conclusion

Within-day comparisons were more reliable than between–day comparisons for both kinematic data and temporal-spatial data. It can be shown from the kinematic graphs that sagittal plane values were less variable than frontal and transverse plane values due to marker application being less variable in the sagittal plane.

The present study revealed acceptable results which were consistent with previous studies for within day and between day repeatability of temporal-spatial and kinematic parameters. This demonstrated that the researcher could reliably place specific protocol markers on subjects on the lower body using the Vicon 370 Kinematic Motion Analysis System.

This pilot study was conducted in order to fulfil objective (d), chapter 5, which investigates lower limb biomechanics in a group of asymptomatic subjects using the kinematic system. It is essential when using this system that variables affecting measurement values need to be eliminated or reduced so confidence can be assured when making clinical decisions regarding treatment based on the results.

Chapter 5 - Analysis of Kinematic and Temporal-Spatial Data of Asymptomatic Subjects using VICON 370 Kinematic Motion Analysis System Whilst Walking

5.1. Introduction

Clinical gait analysis typically seeks to discriminate between what is normal and abnormal gait and to assess change in walking over time (Baker 2006).
It is very important for a health care provider to validly and reliably measure patients' gait variables, as they will be better able to make evidence-based decisions in the clinical management depending on their gait dysfunction. For example, temporal-spatial variables provide a description of a patients' gait against which their previous performance can be compared and can be used to indicate whether improvement or deterioration has taken place. These measures can also be compared against age and sex-matched normal population distributions to ascertain whether the patient is approaching normal performance (Ostrosky et al 1994). It is therefore very important to have a "normally distributed" database of measurements which what to compare the "abnormally distributed" to.

Lower limb biomechanics are thought to be related to many lower limb musculoskeletal conditions (James et al 1978; Tiberio 1987; Tiberio 1988) with anterior knee pain being the most common injury seen in the sports injury clinic (Taunton et al 2002). Knowledge of kinematic differences between individuals with and without AKP is important to health professionals and researchers. This knowledge is needed to develop and optimise valid treatment and prevention strategies (Barton et al 2009).

The main objective of this chapter was to compare kinematic and temporal-spatial data within a group of asymptomatic subjects during barefoot walking using the Vicon 370 Kinematic Motion Analysis System.

5.2. Methodology

This experimental study took place at the Grampian Gait and Motion Analysis Centre, Woodend in Aberdeen. Full ethical approval from the local research and ethics committee was received prior to data collection (project number 01/00057).

Subjects in this study were required to meet the following inclusion and exclusion criteria prior to data collection which was based on those used in other recent AKP studies (Bennell et al 2000).

Inclusion criteria

a) No history of Anterior Knee Pain in the last two years.

b) No history of other injuries to the knee joint such as tears of the menisci, ligament or joint capsule or damage to the articular cartilage (Bennell et al 2000).

c) No history of muscle or tendon ruptures of the lower extremities.

d) No history of recurrent patellar subluxation or dislocation (Bennell et al 2000).

e) No history of surgery of the lower extremity in the last three months (Bennell et al 2000).

Exclusion criteria

a) Pregnancy at time of trial.

b) History of O/A or R/A.

c) History of wearing foot orthoses.

d) History of patellar dislocation or subluxation (Bennell et al 2000).

e) History of surgery of the lower extremity in the last three months (Bennell et al 2000).

Subjects were recruited from The School of Health Sciences at The Robert Gordon University and local rugby and running clubs via posters and letters. Fifteen men and fifteen women with ages ranging between 22 and 49 years (mean 30.21 ± 5.8 years) participated in this study. All subjects were given an information sheet explaining the study and gave written informed consent before taking part.

Using two samples of size 15, detectable differences for each of the variables could be assessed for a two-sample t-test at a 5% significance level and 80% power. For example the power curve below illustrates that a detectable difference of ankle width between genders of 0.53cm would be detectable with an assumed standard deviation of 0.5cm.

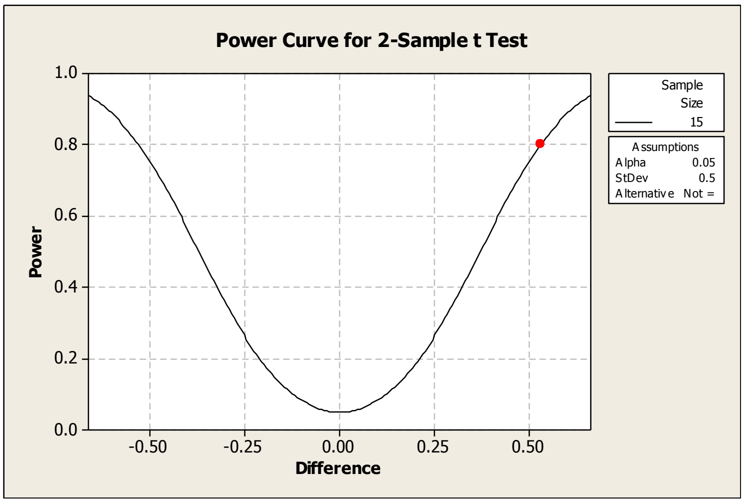


Figure 5.1 Power analysis[19]

This process was repeated with the other variables being studied to include leg length, navicular height, Q-angle supine, hip flexion/extension, knee joint flexion/extension, ankle joint dorsiflexion/plantarflexion, rearfoot angle, RCSP and NCSP.

Upon entering the laboratory, each subject completed a consent form and was asked some short questions which were related to the clinical assessment. These were used to gather information regarding physical activity levels (how often they

[19] SPSS 15 (Chicago, IL)

exercised and which type of exercise did they participate in most often), previous or current lower limb injuries and any previous treatment received. This was done to ensure the subjects were asymptomatic in order to be used as controls. Subjects were asked to wear shorts and remove socks and shoes.

A clinical examination was performed on each subject by the same researcher using the same measurement tools which had been piloted to determine intratester reliability and measurement protocol (chapter 3 for methodology and results).

Active range of motion (ROM) was assessed using a standard double-armed goniometer constructed of clear flexible plastic with a scale of one-degree increments. Lower extremity measurements were examined in both weight bearing and non-weight bearing positions. The clinical measurements carried out under specific protocol for subject position, skin markings and placement of the goniometer and tape measure were as follows:

Non-weight bearing- (supine lying)

1. Hip joint flexion/extension
2. Knee joint flexion/extension
3. Quadriceps angle
4. Leg length measurement
5. Distance ASIS-ASIS (required for kinematic analysis)
6. Ankle joint dorsiflexion with knee flexed at 90 degrees and knee extended
7. 1st metatarsalphalangeal joint dorsiflexion
8. Calcaneal/rearfoot angle (eversion or inversion) and calf-to-calcaneal angle. Calf-to-calcaneal angle was measured with a goniometer as the angle between the vertical midline of the calf and the vertical midline of the calcaneus.

Weight bearing - (relaxed calcaneal stance position)

9. Quadriceps angle

10. Resting calcaneal stance position

11. Neutral calcaneal stance position

12. Navicular height/drop

Visual observations were also conducted by the same tester with the subject standing in their normal stance position. It was noted if the hips were flexed, extended or neutral, knees were flexed, hyperextended or neutral, if there was a presence of genu varum, valgum or recurvatum and the patellar position was also recorded whether it was high, low or normal. Foot type was assessed as to whether it was planus, cavus or normal (Dahle et al 1991). A low arched foot which the MLA was nearly in full contact with the floor was classified as a "planus foot", a high arched foot was classified as a "cavus foot" and the remaining ones were described as a "normal foot". The subtalar joint was noted whether it was in a normal, everted or inverted position in relation to the floor.

Base of gait and angle of gait positions were also sought. The angle of gait is the foot angle made with the line of progression. The base of gait (sometimes called walking base) is the side-to-side distance between the line of the two feet, usually measured at the midpoint of the heel but sometimes the centre of the ankle joint (Whittle 1992). No attempt was made to measure the exact degree of any observable deformities. The lower extremity ranges of motion values selected as "normal" were obtained from previously established normal's (Norkin & White 1985).

Age, gender, height and weight were recorded to determine group homogeneity. Various recommended clinical anthropometric measurements were taken.

a) Height was recorded by a stadiometer
b) Body weight without shoes was measured
c) Knee width was determined from measuring the distance between the lateral and medial femoral condyles in a supine position using a knee

alignment device (KAD) of both knees. These should be measured at the same level as the markers used to identify the flexion/extension axes

d) Ankle width was determined from measuring the transmalleolar distance using the KAD of both ankles. These also should be measured at the same level as the markers used to identify plantarflexion/dorsiflexion axes.

Both the ankle width and knee width positions and measurements are well known and have been thoroughly validated (Davis et al 1991).

5.2.1. Instrumentation

Three-dimensional analysis is the state of the art method used for the analysis of joint kinematics in humans (Deschamps et al 2011).

There are numerous ways of measuring foot biomechanics using cadavers, surface markers and bone pins and each come with their own set of advantages and disadvantages. (See section 2.7 – a review of the literature available on past and present foot models). It is important to take these factors into account as these form the basis for the clinical validation of an assessment tool (Swartz 2004). It is however, generally agreed that skin- mounted markers are the most convenient approach to use in gait laboratories (Robertson et al 2004). It is based on accurate placement of reflective markers over bony landmarks of the pelvis and lower limbs.

Using marker placement sets requires mathematical models to relate measurements of the marker positions to the positions of the lower limb joints. Such models require certain measurements of the subject before gait analysis can take place such as leg length, knee width and ankle width. Any assumptions made regarding gait analysis must be based on accurate placement of these markers. Before the start of the data collection, a test-retest and intrarater reliability study was performed to ensure that variables affecting measurement values were reduced or eliminated by determining the reliability of the tester at marker placement. This was performed by applying the marker set to two subjects independently, three times and on three separate occasions (chapter 4). The results showed acceptable values which were consistent with previous studies which further enhanced the basis of the clinical decision to use the Vicon 370 Motion Analysis System for this study.

The type of data which was derived from the data of each limb of each subject is listed below.

Maximum (max) values were detected for each variable throughout an average cycle. 15 parameters were investigated using the Helen Hayes marker set (Kadaba et al 1987),- ankle dorsiflexion/plantarflexion, abduction/adduction, internal/external rotation, knee flexion/extension, varus/valgus, internal/external rotation, hip flexion/extension, adduction/abduction, internal/external rotation, pelvis anterior/posterior tilt, obliquity up/down and internal/external rotation.

Subject kinematics and temporal spatial data were assessed using the Vicon 370 Kinematic Motion Analysis System (Oxford Metrics, Oxford, UK), two Kistler Force Platforms (Kistler, Switzerland) and Workstation. The graphs were presented by Polygon software. The force platforms were installed flush with the floor in the middle of a 10m walkway. The force platform is normally used to record the vertical (Fz), anteroposterior (Fy) and mediolateral (Fx) components of the external forces and the corresponding moments (Nadeau et al 1997) however these were not measured in this present study. The position of the force plates on the floor was not revealed to the subjects in order to eliminate targeting resulting in altered gait patterns (Heiderscheidt 2002).

Six infrared cameras operating at a 50 HZ sampling rate were positioned such that they could detect all of the twenty-six markers simultaneously during the whole gait cycle.

Kinematic data was calculated using the standard Helen Hayes lower body marker set (Kadaba et al 1987). Small, light reflective spherical markers (approx. 2cm in diameter) were attached to various anatomic landmarks on each subject with hypoallergenic double-sided tape. Marker placement is known to be critically important in the measurement process and was described in some detail in chapter 4.

The markers were located according to the specifications in chapter 4

SACR	**Sacrum**
LASI/RASI	**Left anterior superior iliac spine/Right anterior superior iliac spine**
LTHI/RTHI	**Left lateral thigh/Right lateral thigh**
LKNE/RKNE	**Left lateral condyles of knee/Right lateral condyles of knee**
LTIB/RTIB	**Left lateral lower 2/3 of tibia/Right lateral lower 2/3 of tibia**
LHEE/RHEE	**Left heel/Right heel**
LANK/RANK	**Left lateral malleoli/Right lateral malleoli**
LTOE/RTOE	**Left 2nd ray/Right 2nd ray**

Table 5.1 Anatomical location of markers

From the kinematic data, the positions of the hip, knee and ankle joint, in the sagittal, frontal and transverse planes were calculated with Workstation and Polygon software. These values were calculated separately for each extremity since the software averages the lengths of the limbs as a scaling parameter when the location of the centre of the hip joint is determined (Eng & Winter 1995). All participants were assigned an identification number to preserve confidentiality.

5.2.2. Statistical analysis

All statistical comparisons were analysed using the Statistical Package for Social Sciences 15 (SPSS Inc, Chicago, IL). Before running any tests, the data was checked for anomalies such as extreme values or skewed distribution. This was done by running a normality test (Shapiro Wilks) and constructing scatterplot graphs. This ensured that there was a normative database in which to test against the symptomatic group (chapter 6).

Means and standard deviations were obtained for the clinical assessment, temporal-spatial and kinematic data. Independent sample *t*-tests were used in the clinical assessment and temporal-spatial data to evaluate group differences between right and left values and gender comparisons were also compared to establish if there were any differences between the right and left sides of the body. All were tested at a two-tailed significance of ($p<0.05$).

Comparisons were made between subjects for stride characteristics (velocity, cadence and stride length), percentage of gait cycle in single and double limb stance, force plate parameters and knee, foot and leg kinematics. Three clinical planes of motion (sagittal, frontal and transverse) were analysed at the pelvis, hip, knee, ankle and foot joints. All joint comparisons were examined bilaterally. For each kinematic variable, the mean of five trials were used in the analysis.

5.3. Results

Due to the enormity of results analysed, significant results in the form of graphs and tables are shown in the results section but a full and concise list of all results can be found in the appendix. Normality tests (Shapiro Wilks) were performed to check for outliers and to ensure we had a normative database in which to test against the symptomatic group. All variables plotted produced a normal linear line with the exception of opposite foot contact and opposite foot off variables. These two contained two outliers which skewed the data negatively. An outlier is an observation which lies an abnormal distance from other values in a random sample from a population. Interestingly it was the same subject which skewed both variables. This was attributed to a measurement error and the data was then tested again without this particular subject and a normal bell shaped curve was produced.

Descriptive statistics of the symptomatic subjects who participated in the study are shown in the tables below.

5.3.1. Normal subject demographics

Fifteen males and fifteen females were recruited with a mean age of 29.4 years, 182cm in height for the males and 163cm for females and 80.3kg in weight for males and 68.2kg for females.

	N (no.)	**Min**	**Max**	**Mean**
age	30	22	49	29.43
Height (cm)	30	153	198	173.0
Weight (kg)	30	51	100	74.26

Table 5.2 Subject demographics, age height and weight

gender		**N (no.)**	**Mean**	**SD**	**SEM**
Height (cm)	**male**	15	182.4	8.1	2.0
	fem	15	163.7	6.2	1.6
Weight (kg)	**male**	15	80.3	10.1	2.6
	fem	15	68.2	15.0	3.8

Table 5.3 Subject demographics, height and weight in males and females

An independent t-test revealed a statistically significant difference between height and weight (height, $t(28) = 7.04$, $p = 0.001$; weight, $t(28) = 2.58$, $p=0.015$).

5.3.2. Clinical assessment

A matched paired sample *t*-test was conducted to test whether there were any differences between right and left values in the normal group. It was proven that there was no significance between any of the variables in right and left sides and shows that there is general tendency for symmetry. Due to this, left and right values have been averaged together.

Influence of gender on asymptomatic group

An independent *t*-test was used to test for a difference between gender and right and left measurements averaged together.

Variable	Gender	Mean	std.dev	p- value
Ankle width (cm)	**Male**	7.52	0.46	0.0001
	Female	6.84	0.39	
Leg length (cm)	**Male**	98.20	4.77	0.0001
	Female	89.05	3.80	
Navicular height (mm)	**Male**	39.50	7.20	0.014
	Female	33.75	5.33	
Q-angle supine (°)	**Male**	9.60	0.85	0.0001
	Female	14.97	3.20	
Q-angle weight bearing (°)	**Male**	10.57	1.71	0.001
	female	14.90	3.75	

Table 5.4 Variables which showed significant differences when comparing influence of gender

There were five variables which showed significant difference between males and females in the normal group (Ankle width ($p= 0.0001$), leg length ($p= 0.0001$), navicular height ($p=0.014$), Q-angle weight bearing ($p=0.001$) and Q-angle supine ($p=0.0001$)).

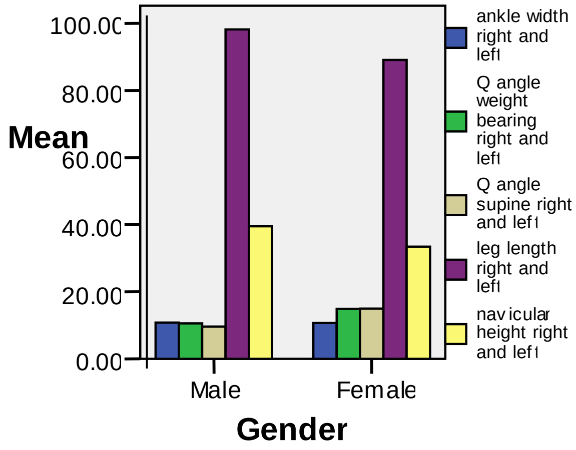


Figure 5.2 Variables which show statistical significance between genders (ankle width (cm), Q-angle (degrees), leg length (cm) and navicular height (mm).

5.3.3. Temporal spatial parameters

The temporal spatial data was analysed and shown in the table below

VARIABLE	**MEAN**	**SD**
CADENCE (steps/min)	119.7	8.65
DOUBLE SUPPORT (s)	0.20	0.03
FOOT OFF (%)	59.99	1.32
OPPOSITE FOOT CONTACT (%)	49.91	1.09
OPPOSITE FOOT OFF (%)	10.04	1.86
SINGLE SUPPORT (s)	0.39	0.03
STEP LENGTH (m)	0.70	0.07
STEP TIME (s)	0.50	0.04
STRIDE LENGTH (m)	1.42	0.12
STRIDE TIME (s)	1.01	0.07
WALKING SPEED (m/sec)	1.39	0.16

Table 5.5 Temporal spatial data from normal's (n=30)

Note-right and left values have been averaged together

An independent *t*–test found significant gender differences for cadence, double support, step time and stride time ($p<0.05$).

Normality tests were conducted and all of the variables were normally distributed with the exception of cadence. The scatterplot graph (figure 5.3 and 5.4) shows an extreme value which negatively skews the data.

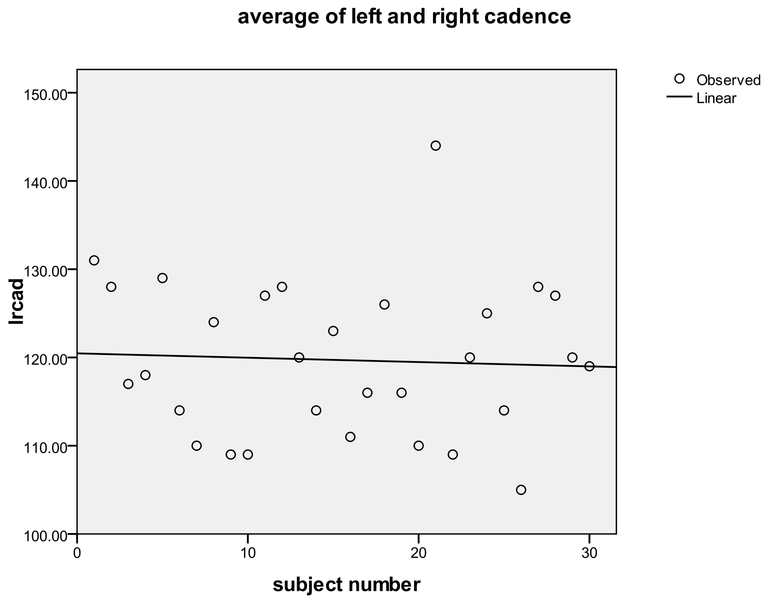


Figure 5.3 Curve estimation of right and left cadence (steps/min)

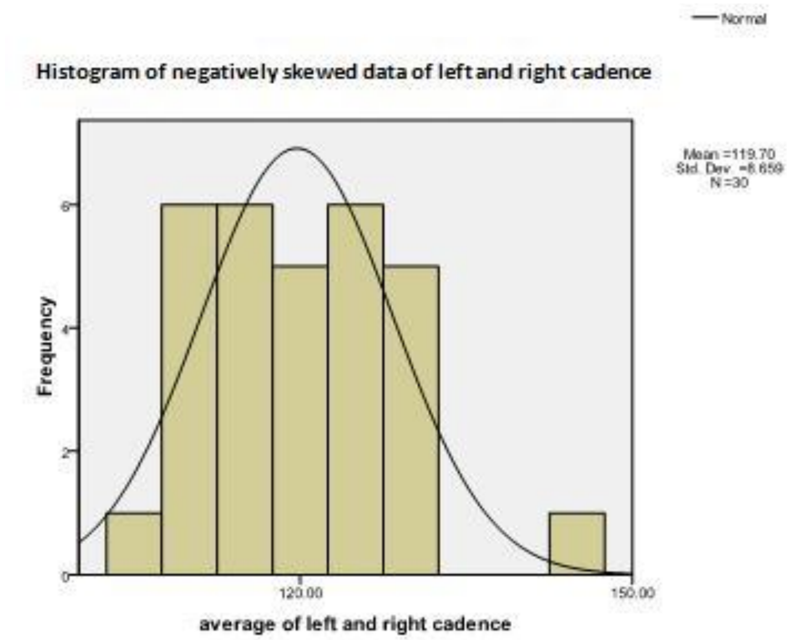


Figure 5.4 Histogram of skewed data of left and right cadence (steps/min)

5.3.4. Kinematic data analysis

The power analysis conducted on the kinematic variables suggested that there was a 93% chance of detecting any differences in these variables between the test conditions which were greater than 3.5 degrees. An independent *t*-test was carried out to test for any differences in the test variables between males and females. The results are divided into maximum and minimum values in the averaged gait trial.

Comparison of maximum values between male and female asymptomatic subjects

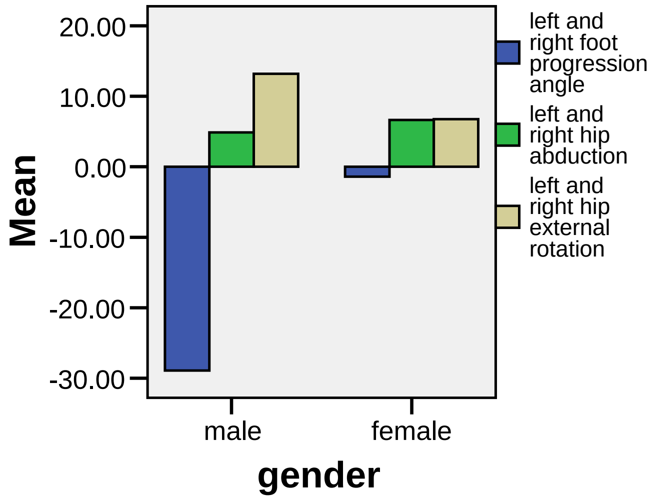


Figure 5.5 Significant maximum variables between males and females in asymptomatic barefoot walking (N=30). Angles are measured in degrees.

Variable control	**Male (°) mean**	**Fem (°) mean**	***P* value**
Foot progression	-28.9	-1.4	0.01
Hip abduction	4.86	6.63	0.027
Hip external rotation	13.18	6.76	0.045

Table 5.6 Values for foot progression, hip abduction and hip external rotation

Three kinematic variables showed significant differences under the influence of gender. The table shows a large increase in foot progression angle between men and women and a much smaller decrease in angle for hip external rotation in women.

Normality tests, to test for skewness were carried out in order to determine that the control group was a normal representitive sample of the population. This was required to have a normal database for which to test the symptomatic subjects with. The graphs below show the normal data which has been negatively skewed (-3.76).

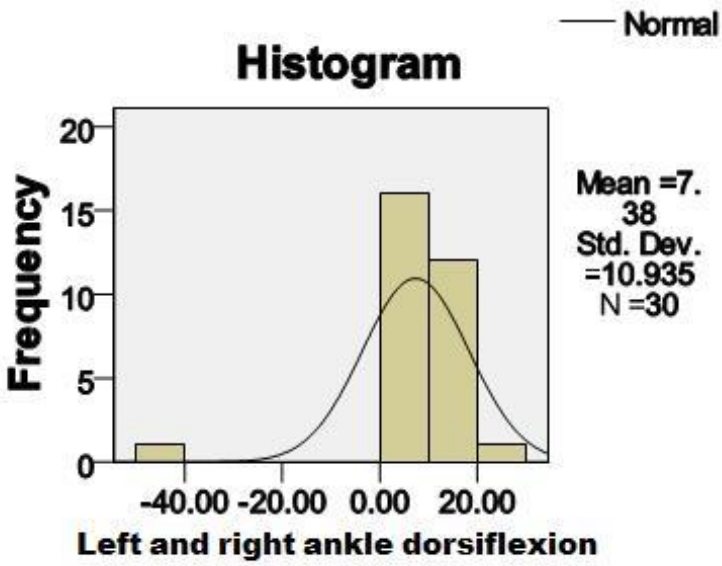


Figure 5.6 Histogram of negatively skewed data in left and right ankle joint dorsiflexion (°)

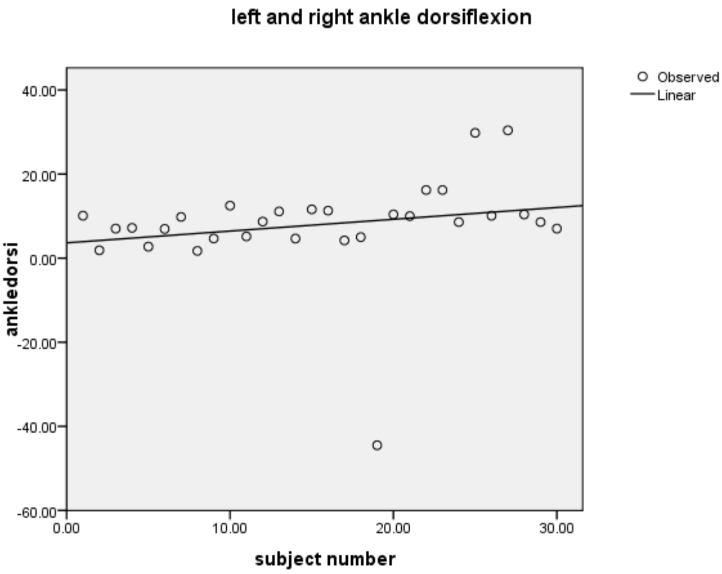


Figure 5.7 Curve estimation of right and left ankle dorsiflexion (°)

Test of normality (Shapiro-Wilks) show significance for this variable within the thirty subjects ($p<0.05$). It can be seen from the histogram that there is an outlier with a value of -40 degrees. However the case with the outlier had extreme values throughout all the variables and it was decided it must be a computer error as this value couldn't be a clinical value. The subjects original clinical raw data was checked and it was within normal ranges however a problem must have occurred during the kinematic process such as marker placement error, marker wobble or simply a cumputer error. For these reasons, subject 19 was removed and normality tests were carried out again.

The graphs below show the data without the extreme value and although there are a couple of larger numbers, the data is almost sitting on the line which is what we would expect with a normal sample of subjects. Skewness did improve without number 19 and the value was nearer zero which is what we expected (0.882). Figures 5.6 and 5.7 illustrates normal values when subject 19 has been omitted.

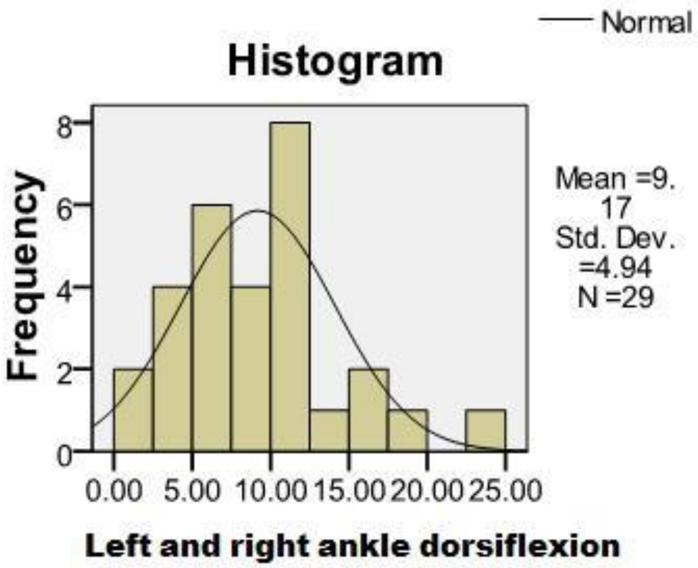


Figure 5.8 Histogram diagram of right and left dorsiflexion without subject 19 (°)

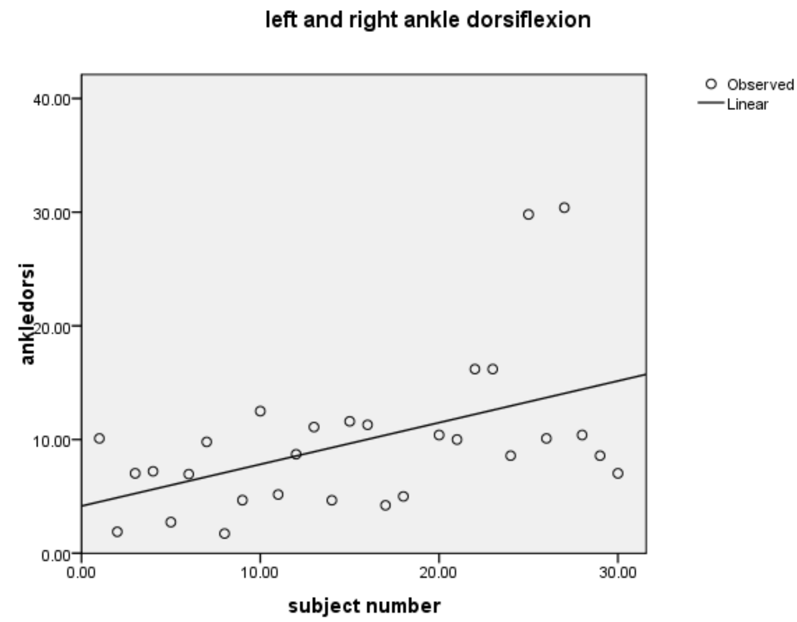


Figure 5.9 Curve estimation of left and right dorsiflexion without subject 19 (°)

All the other variables were tested and all were not significant and positively skewed with values close to zero.

It was accepted following inspection of the raw data that the reason for the outlier from subject 19 was a computer error so the same tests were run again and no outlier was evident on this occasion therefore normal values were found. Because of this the kinematic graphs showed normal wave patterns and a full concise set of normal graphs can be found in the Appendix.

5.4. Discussion

The main objective for this study was to evaluate the normal effects of walking barefoot by analysing the kinematic data and the temporal parameters. This was conducted in order to achieve a normative database in which to compare the symptomatic data with.

The success of this study relied on the ability to accurately locate the reflective markers worn by the subjects and to carry out a reliable clinical assessment on each individual. This was achieved in two pilot studies conducted previously where a high reliability score was achieved. Intraclass correlation coefficients ranged from (ICC= 0.37 to 0.98) indicating from the values stated above acceptable reliability for these measurements.

A common method of finding a normal foot, knee or hip parameter such as normal Q-angle or normal dorsiflexion of the ankle according to Phillips (2000) is to examine a group of young asymptomatic adults and find the statistical average for whatever morphologic variable is being analysed. A standard deviation from the average is determined followed by a statement that anyone who falls within two standard deviations should be considered normal. The results demonstrated this and therefore could be considered as normal.

5.4.1. Clinical assessment

Ball & Johnson (1996) demonstrated a small but significant gender difference with females exhibiting a greater range of motion than males in all the age groups. It was found in this study that only five of the variables had a significant difference between the sexes.

As expected, Q-angle supine and Q-angle weight-bearing were highly significant along with ankle width, leg length and navicular height. The values for all the variables were all larger in the male group with the exception of both Q-angles which supports the literature available that women have larger Q-angle values (Livingston and Mandigo 1999; Emami et al 2007).

Q-angles supine and weight bearing (w/b) values were established (mean male supine (9.6°) and w/b (10.57°); mean female supine (14.96°) and w/b (14.99°) and were consistent with some studies (Horton & Hall 1989). The mean values for both men and women were lower than the normal values suggested by Kantaras et al (2001) which were that normal value is 10 to 15 degrees for men and 15 to 20 degrees for women. The Manual of Orthopaedic Surgery (1972) considers angles greater than 15 degrees in men and 20 degrees in women pathological and values below these values normal. Horton & Hall (1989) found similar Q-angles of 11.2° for men and 15.8° degrees for women and attributed the lower values to the fact that they measured in a weight bearing position as opposed to Agletti et al (1983) who found higher values of 14 and 17 degrees for men and women and measured in a supine position. The results of the present study do not agree with this theory as the values for weight bearing are higher than the supine position.

Navicular height was 6mm higher in the male group than the female group. Nielsen et al (2009) found that gender has an effect on navicular drop values although their study was dynamic but they reported a higher male mean value than the female group. They also reported that as the foot length measurement increased by 10mm, navicular drop also increased in height by 0.4mm in males and 0.31mm in females. Although this study did not measure foot length, this may account for the fact that the male group had higher values than the female group as males generally have larger feet than women.

Ankle width and leg length were also significantly different. A bivariate correlation was conducted to test for a relationship between leg length and height and a very strong positive relationship exists ($r=0.95$).

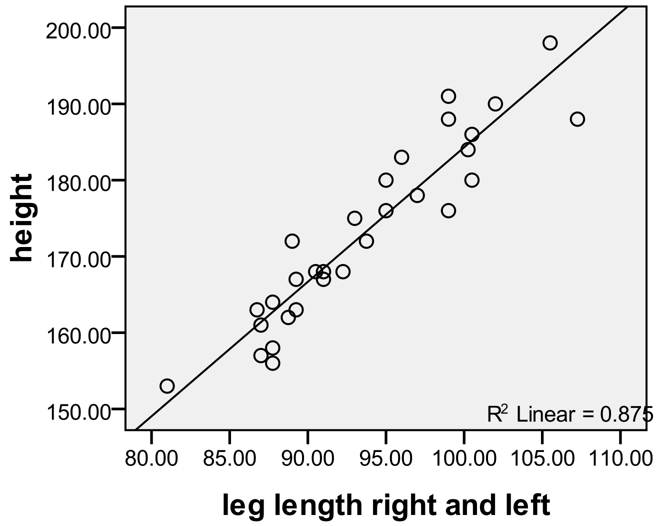


Figure 5.10 Relationship between leg length and height (cm)

From the results it can be seen that the majority of men were taller and therefore had longer legs in the normal group. Ankle width was slightly related with height (r=0.75) which is consistent with significant values throughout the normal group. There was no significance between height and weight measurements in the group.

Boone & Azen (1979) conducted a study which found that the amplitudes of motion of the left and right joints were consistently similar when testing healthy individuals. They concluded that the motions of the joints of a patient's healthy leg can routinely be used for comparison with those of the affected side in the presence of injury.

The results of the clinical assessment show a range of normal values from thirty asymptomatic subjects which is representative of a normal population. It is therefore a justifiable database on which to compare symptomatic subjects against.

5.4.2. Temporal spatial data

This data appeared to provide normal mean values with which any pathological data could be compared against. When compared with gender, four variables showed significant differences. However, because of there being distinct gender differences in skeletal dimensions, the walking patterns of the two genders are likely to differ (Cho et al 2004).

Correlations were tested for cadences, step time and stride time between height and leg length. It was expected that cadence would be significant between males and females as they are generally taller and have longer leg length however no relationship was shown. It is also a common observation that smaller people tend to walk with smaller steps but at a higher step time than taller people do (Hof 1996) and this would be consistent with this study's results (cadence for males, 115 steps/min and females, 123 steps/min and step time for males 0.52s and females 0.48s). Crosbie et al (1997) found that females demonstrated a higher cadence value than males. The female cadence at free speed was around 6% higher than the males. Oberg et al (1993) studied what the influence of gender was on normal women and men and found that step timing was higher and step length and walking speed was lower for women than for men. Their results are in accordance with the present study's results.

Regression lines are shown to investigate the relationship between cadence, leg length and height.

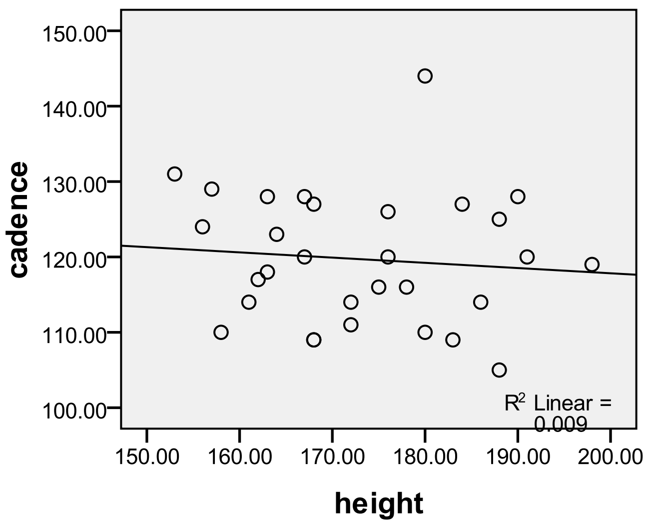


Figure 5.11 Linear regression line describing cadence (steps/min) and height (cm)

This graph shows no relationship between the two (r=-0.095).

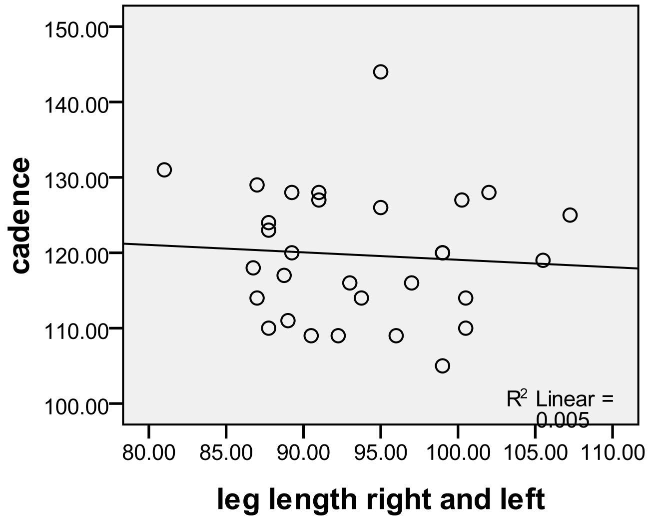


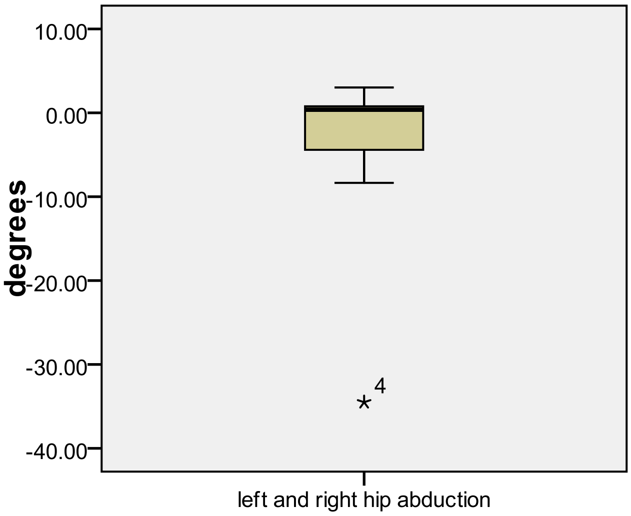
Figure 5.12 Linear regression line describing cadence (steps/min) vs. leg length (cm)

The graph above again shows no relationship (r=-0.072).

5.4.3. Kinematic data

It was noted when looking at the waveforms of the kinematic graphs that there were a couple of unexpected patterns within the range of the graph. This occurred in many of the different variables so an extreme values test was carried out to look for outliers. In this group of asymptomatic subjects there were two specific subjects who continuously had extreme values whether it was the highest or the lowest. This is shown in the box plots below.

Minimum boxplots



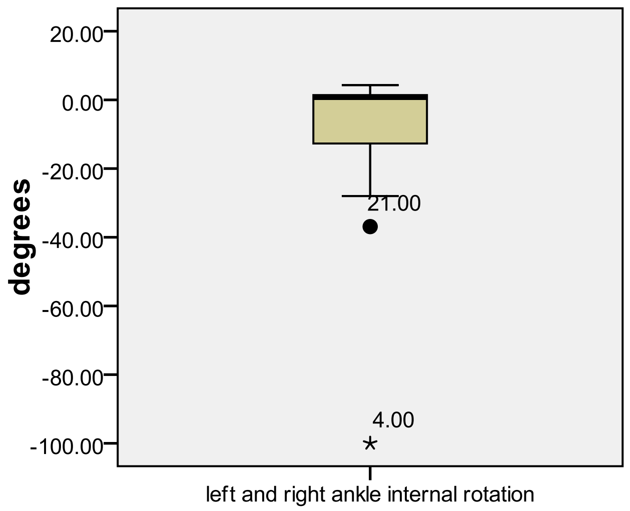


Figure 5.13 Minimum boxplots of normal group for hip abduction and ankle internal rotation

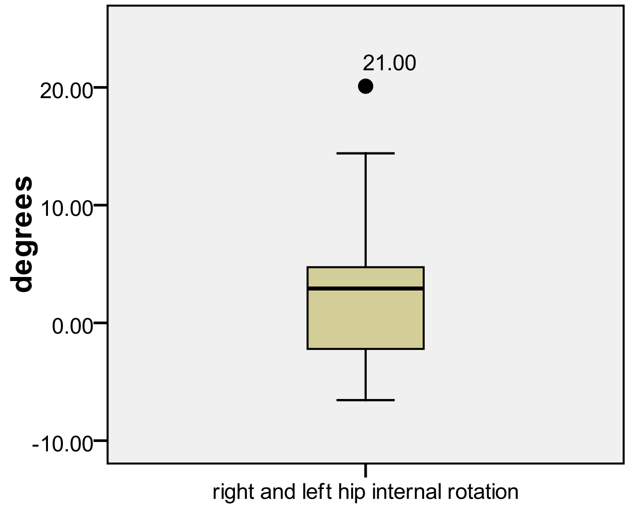


Figure 5.14 Maximum boxplot of normal group for internal hip rotation

As can be shown from the graphs above, subject 4 and 21 had outlying values within these three variables. Subject 4 and 21 in ankle internal rotation and hip abduction were extreme outliers where the values were more than three box widths away from the rest of the data. However, subject 21 in hip internal rotation variable showed only a mild outlier which was within the 1.5 box width range. Outliers have to be investigated carefully. They can contain certain information about the process under investigation or the data collection and recording process. The root cause of these outliers has to be determined as to why they appeared so a re-test can be justified. In these cases, it was concluded that due to the extreme values, it was nothing to do with the participant structurally but perhaps to do with recording error, imputing error, marker placement error, marker wobble or simply a computer processing error. On review of the data, it was decided that because the cause of the extreme values could not be identified, the data was run again and normally distributed data was found within these three variables.

5.5. Conclusion

The main aim of this chapter was to investigate and establish values for lower limb biomechanics in a group of normal individuals using a detailed clinical assessment and the Vicon 370 Kinematic Motion Analysis System. Significant differences were found between genders but this was expected as there are distinct gender differences in skeletal dimensions, the walking patterns of the two genders are likely to differ (Cho et al 2004).

One of the reasons for this objective was to obtain an asymptomatic database in which it was possible to compare symptomatic subjects with. The normal set of results which was obtained from this chapter can be used effectively against the set of results from symptomatic subjects in the next chapter.

Chapter 6 – Analysis of Kinematic and Temporal-Spatial Data of Symptomatic Subjects using Vicon 370 Kinematic Motion Analysis System whilst Walking under Three Experimental Conditions.

6.1. Introduction

The aim of this study was to compare barefoot trials with shod and medially posted orthotic devices to establish any changes in foot and leg mechanics using 3D motion capture, in this instance the Oxford Metrics Vicon 370 Kinematic Motion Analysis System.

6.2. Methodology

The methodology for the symptomatic group differed in that this experimental study took place at The Robert Gordon University Gait Laboratory in Aberdeen. Due to new ownership and relocation of the laboratory this study was conducted in different premises to the asymptomatic study. However, every effort was made to ensure the experimental conditions were made the same. The Bioengineer, the researcher and the equipment were constant variables.

The inclusion and exclusion criterion that was used for recruiting the normal group of subjects was slightly adapted to recruit symptomatic subjects.

Inclusion criteria

a) Been diagnosed as having AKP by a medical doctor/Physiotherapist (Bennell et al 2000 and Ng & Cheng 2002).

b) Having anterior or retropatellar knee pain (Bennell et al 2000; Cowan et al 2001).

c) Reported that at least two of the following activities exacerbated their symptoms

- prolonged sitting

- climbing up and down stairs

- kneeling

-squatting

(Bennell et al 2000; Watson et al 2001; Thomee et al 2002 and Cowan et al 2002).

Exclusion criteria

a) History of surgery of the lower extremity in the last three months (Bennell et al 2000).

b) Pregnancy at time of trial.

c) History of wearing orthoses.

d) History of patellar dislocation or subluxation (Bennell et al 2000).

e) History of O/A or R/A.

Subjects were recruited from The School of Health Sciences at The Robert Gordon University, local rugby and running clubs via posters and flyers. Full ethical approval from the local research and ethics committee was received prior to data collection (project number 01/00057).

Fifteen subjects (7 men and 8 women) with ages ranging between 22 and 49 years (mean 30.21 ± 5.8 years) participated in this study and met with the appropriate criteria. All subjects were given an information sheet explaining the study and gave written informed consent before participating in the study.

Detectable differences were assessed for paired *t*-tests at 5% significance levels and 80% power for each of the variables using a sample size 15. The variables assessed were the same as tested in chapter 5- hip flexion/extension, knee flexion/extension, Q-angle, leg length, ankle dorsiflexion/plantarflexion, 1st MTPJ dorsiflexion, rearfoot angle, navicular height, NCSP and RCSP.

The subjects were given similar questions to the asymptomatic group and this was used in the same way to categorise this group of subjects as experiencing unilateral knee pain or bilateral knee pain. Subjects were asked to wear shorts and remove socks and shoes.

Age, gender, height and weight were recorded to determine group homogeneity. The same recommended clinical anthropometric measurements as the control group were taken.

An identical clinical assessment to the one used on the control group (see chapter 4) was performed on each pathological subject using the same measurement tools for each one. The protocol was exactly the same between the two groups and executed under identical experimental conditions.

In each experimental condition, subjects walked at a self-selected speed. Each subject walked under three experimental conditions

a) Walking barefoot

b) Walking with their own trainers

c) Walking with their own trainers and supplied orthoses

The insole of the trainer was removed for the third experimental condition and replaced with an orthotic device to allow the shoes to fit. Each device was fitted corresponding to shoe size.

The orthotic device was selected because it was easily available to the general population, inexpensive and a good example of the many "off the shelf orthotics" for the subject experiencing AKP.

The information below describes the orthotic in detail and explains exactly what the device is designed to do.

Figure 6.2 ORTHAHEEL Regular designed by podiatrist Phillip Vasyli

a) Designed to fit regular men's and women's (flat) shoes.

b) Made of flexible E.V.A (similar to the material used for midsoles in most sports shoes).

c) Supporting the foot around its neutral position via a 4 degree built-in rearfoot varus angle.

d) Stabilizing the foot via an Angular Restraining Mechanism. This ARM is a trademark system and controls frontal plane motion (calcaneal eversion) (Vasyli Inc).

e) Correctly aligning the 2nd, 3rd and 4th metatarsals.

f) Shock-absorption with a Shock Dot in the centre of the heel.

g) Forefoot balancing with a 4 degree wedge.

The expected outcome of using the orthotic was to see a decrease in eversion variables between heel strike and midstance. A further expectation was that due to the coupling mechanism at the ankle, internal rotation of the leg would be reduced and, a third outcome was that external rotation of the tibia would be reduced between midstance and toe-off (Stacoff et al 2007).

The orthotic devices were utilised to test if they reduce abnormal pronation in gait as measured by the calf-to calcaneus or the calcaneus-to-vertical angle. Sims (1983) and Novick & Kelly (1990) both found these angles reduced in subjects walking wearing orthoses compared with subjects walking with shoes alone.

6.2.1. Hypotheses

The expected effect of comparing the barefoot trials with the shod and medially posted orthotic trials would be to see a reduction of foot and leg variables. It was also expected that due to the coupling mechanism at the ankle, internal tibial rotation and foot eversion would be reduced when wearing the orthotic (Stacoff 2007).

A further 3 parameters were also evaluated as a result of the extra markers on the foot and lower leg.

a) 3D calcaneal inversion/eversion - referred to as pronation/supination.

b) EV/TIR relationship - (discussed in chapter 2.3)

c) Navicular height

6.2.2. Marker placement

Subject kinematic and temporal spatial data were assessed using the same 3D measurement tool as the control group (chapter 5).

The markers from the Helen Hayes marker set were located according to the specifications below. The additional markers on the foot model are also listed. (These have been discussed in detail in a previous chapter 4.4).

SACR	**Sacrum**
LASI/RASI	**Left anterior superior iliac spine/Right anterior superior iliac spine**
LTHI/RTHI	**Left lateral thigh/Right lateral thigh**
LKNE/RKNE	**Left lateral condyles of knee/Right lateral condyles of knee**
LTIB/RTIB	**Left lateral lower 2/3 of tibia/Right lateral lower 2/3 of tibia**
LHEE/RHEE	**Left heel/Right heel**
LANK/RANK	**Left lateral malleoli/Right lateral malleoli**
LTOE/RTOE	**Left 2nd ray/Right 2nd ray**
LNAV/RNAV	**Left navicular tuberosity/Right navicular tuberosity**
LCALC/RCALC	**Left calcaneal marker/Right calcaneal marker**
LCALC 1/ RCALC 1	**Left calcaneal marker 1/ Right calcaneal marker 1**
LCALC 2/RCALC 2	**Left calf marker 2/ Right calf marker 2**
LCALC 3/RCALC 3	**Left distal calf marker 3/ Right distal calf marker 3**

Table 6.1 Locations of the anatomical landmarks

From the kinematic data, the positions of the hip, knee and ankle joint in the sagittal, frontal and transverse planes were calculated with Vicon workstation.

These values were calculated separately for each extremity since the software averages the lengths of the limbs as a scaling parameter when the location of the centre of the hip joint is determined (Eng & Winter 1995). Previous studies have measured the ROM over the entire gait cycle but due to the STJ working in different ways throughout the cycle it was decided to only examine the effect during the contact phase (heel contact to foot flat), mid-stance (foot flat to heel-off) and propulsion (heel-off to toe-off). The non-weight-bearing swing phase was not examined. Graphs were presented with Polygon software. All participants were assigned an identification number to preserve confidentiality.

6.2.3. Statistical analysis

All statistical comparisons analysed using the Statistical Package for Social Sciences 15 (SPSS Inc, Chicago, IL) and consisted of descriptive statistics, independent *t*-tests and matched paired sample *t*-tests. All were tested at a two-tailed significance of ($p<0.05$).

Comparisons were made between subjects for stride characteristics (velocity, cadence and stride length), percentage of gait cycle in single and double limb stance, force plate parameters and foot and leg kinematics. All subjects were tested for normality using the Shapiro-Wilks test.

A repeated measure one-way ANOVA analysis was performed to compare the minimum and maximum angle values throughout various points of the gait cycle within the three experimental conditions (barefoot, trainers and trainers with orthoses). In comparing the results obtained under the three test conditions, the difference between equivalent measurements was deemed to be significant if the corresponding P value was less than 0.05. Pairwise comparisons were also sought as although the results may show a statistically significant result which suggests that there is a difference somewhere among the groups. However, this does not tell you which group differs from one another. Where significance was found, the Bonferroni post hoc test was executed to identify where the difference lay and the assumption of Sphericity was assessed using Mauchlys test. All data were compared to previously obtained results on a group of asymptomatic subjects using various *t*-tests (chapter 5).

The power analysis conducted on the kinematic variables suggested that there was a 93% chance of detecting any differences in these variables between the test conditions which were greater than 3.5 degrees. The kinematic variables were the same variables as tested in chapter 5.

6.3. Results

Descriptive statistics of the subjects who participated in the study are shown in the tables below.

6.3.1. Subject demographics

	N	Minimum	Maximum	Mean	Std. Deviation
Age	15	27	49	36.53	7.219
Height (cm)	15	160.50	187.50	173.46	8.578
Weight (kg)	15	49.00	105.00	77.00	16.392

Table 6.2 Subject demographics for age, height and weight

	gender	N	Mean	Std. Deviation	Std. Error Mean
Height (cm)	**male**	7	180.35	6.811	2.574
	female	8	167.43	4.288	1.516
Weight (kg)	**male**	7	89.00	12.110	4.577
	female	8	66.50	11.904	4.208

Table 6.3 Subject demographics for height and weight between males and females

An independent *t*-test was conducted to compare the height and weight differences in men and women. There were no significant differences difference between, height and weight ($p<0.05$).

6.4. Clinical Assessment results

A matched paired *t*-test was conducted to test for any significance between the pathological leg and the unaffected leg in the patient group. It showed a statistically significant difference between right and left sides when measuring Q-angle weight bearing ($p=0.048$), rearfoot angle ($p=0.009$) and hip flexion range of motion ($p=0.023$). It was, however expected to see more of the variables showing significance but all the other variables showed no difference between right and left ($p>0.05$).

Variable	**Side of body**	**Mean**	**Std. deviation**	**P value (<0.05)**
Q-angle weight bearing (°)	**Left**	12.86	4.77	0.048
	Right	13.33	4.87	
Rearfoot angle (°)	**Left**	8.46	7.68	0.009
	Right	9.66	8.73	
Hip flexion ROM (°)	**Left**	116.13	9.15	0.023
	right	114.13	10.15	

Table 6.4 Variables that differed significantly between right and left sides of the body

The graph below displays results when comparing the normal group with 30 unaffected legs and the symptomatic group with 9 right and 6 left affected legs. Note the similarity in the patterns between the groups.

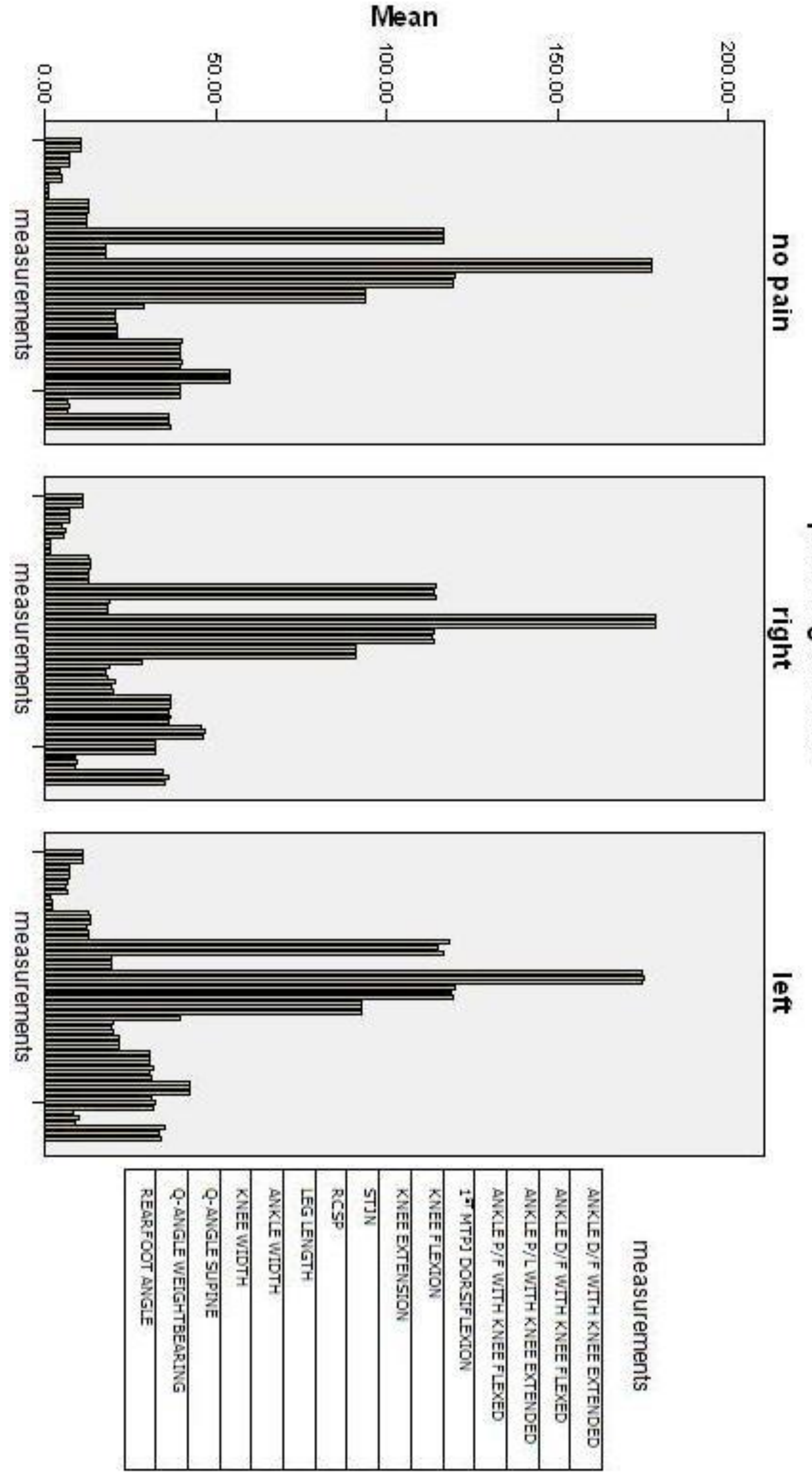


Figure 6.3 Comparisons of normal group with unaffected knees with patient group with affected knees (All angles measured in degrees, leg length, ankle and knee width measured in cm).

6.4.1. Influence of gender on symptomatic group

When testing for the effect of gender, the same independent *t*-test was conducted in the symptomatic group and the same four variables as in the control group were statistically significant. Ankle width right, left and averaged ($p=0.001$, 0.011, 0.003); Q-angle weight bearing right, left and averaged ($p=0.001$, 0.0001, 0.0001); Q-angle supine right, left and averaged ($p=0.011$, 0.001, 0.0001), however only the right navicular height was significant ($p=0.038$) and the right leg length ($p=0.048$).

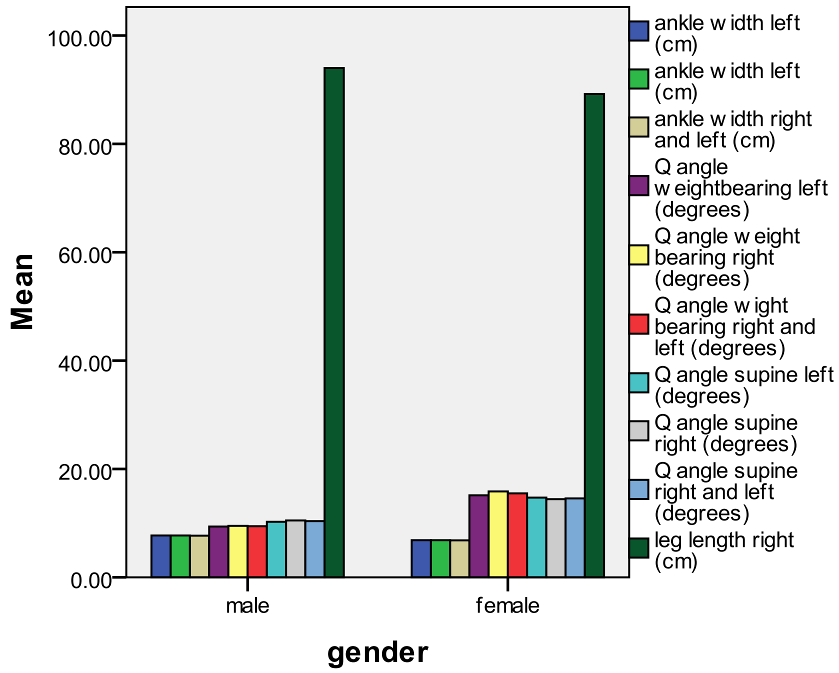


Figure 6.4 Bar chart displaying the variables significantly affected by gender (symptomatic group)

Influence of gender on symptomatic compared with symptomatic group

By using the normal database collected in an earlier chapter (chapter 5), a gender comparison was carried out between this group and the pathological group using an independent *t*-test. Results showed significant differences between males and females for Q-angle supine and weight bearing values and navicular height values ($p<0.05$).

The table below shows the differences between the weight bearing Q-angle and supine Q-angle in both males and females.

Control group (N = 30)

	gender	N	Mean	Std. Dev	Std. Error Mean
Q angle supine right and left (°)	**male**	15	9.6	.84	.21
	female	15	14.9	3.1	.82
Q angle w/ bearing right and left (°)	**male**	15	10.5	1.7	.44
	female	15	14.9	3.7	.96

Injured group (N=15)

	gender	N	Mean	Std. Dev	Std. Error Mean
Q angle w/bearing left (°)	**male**	7	9.0	.69	.26
	female	8	16.7	3.8	1.3
Q angle w/t bearing right (°)	**male**	7	9.1	1.4	.55
	female	8	17.0	3.5	1.2
Q angle supine left (°)	**male**	7	9.2	1.5	.59
	female	8	15.6	2.6	.92
Q angle supine right (°)	**male**	7	9.4	1.2	.48
	female	8	15.7	3.8	1.3

Table 6.5 Differences between weight-bearing Q-angle and supine Q-angle in symptomatic and asymptomatic groups

When looking at the males and females in their respective groups of symptomatic and asymptomatic, it was interesting to note that the female angles were much higher in the symptomatic group compared to the asymptomatic group (mean 14.9° and 14.9° increased to 16.7° and 15.6°) but the male values were lower in the symptomatic group compared with the asymptomatic group (mean 9.6° and 10.5° decreased to 9.0° and 9.2°).

Male values for navicular height increased slightly between the normal and patient groups and female values increased by 3mm between groups.

Differences in clinical assessment between symptomatic and asymptomatic groups

An independent *t*-test was conducted to see any differences between the variables in the symptomatic group and the asymptomatic group. Right and left legs were compared due to the clinical significance of a few variables when comparing both sides.

There was no statistical significance for most of the variables however as expected some of the variables were affected. Rearfoot angle left ($p=0.04$), rearfoot angle right ($p=0.01$), ankle plantarflexion with knee extended in the right ($p=0.03$), and with knee flexed in the left side ($p=0.04$) and 1st MTPJ dorsiflexion ($p= 0.01$) all had statistical significance ($p<0.05$). See Table 6.6.

variable	mean normal	mean patient	Std dev normal	Std dev patient	P value (p<0.05)
rearfoot angle (°) left	6.66	8.46	2.97	2.44	0.04
rearfoot angle (°) right	7.03	9.66	3.38	3.06	0.01
ankle p/flexion knee ext(°)	40.0	34.1	7.66	10.37	0.03
ankle p/flexion knee flexed(°)	39.86	34.2	8.25	9.3	0.04
1st MTPJ d/flexion (°) right	39.6	32.2	7.73	3.27	0.01

Table 6.6 Displaying the variables which showed significance between the two groups

6.4.2. Temporal and stride parameters for symptomatic group (n=15)

A matched paired t-test was conducted to test if there was any significance between the patient's pathological leg and their non-pathological leg. When comparing the spatio-temporal parameters, foot off in the barefoot condition showed significance (left leg mean = 60.68%; right leg mean = 61.78%, p=0.04) between right and left sides but all the other variables showed very small differences in the means but no significant differences. No significant differences were shown in the trainer condition and in the orthotics condition, cadence showed significance (mean 0.95 steps/min, p=0.045).

A one-way repeated measures ANOVA was also conducted to compare continuous temporal spatial variables over the three conditions, barefoot, trainers and orthotics. The means and SD are shown in Table 6.7.

MEASURE	BAREFOOT	TRAINERS	ORTHOSES
Cadence (steps/min)	119 ± 11.31	118 ± 12.0	118.67 ± 11.8
Double support (s)	0.23 ± 0.50	0.25 ± 0.09	0.24 ± 0.07
Foot off (%)	61.1± 1.85	61.2 ± 2.42	60.0 ± 4.15
Opposite foot contact (%)	50.64 ± 2.6	50.41 ± 3.27	51.17 ± 2.80
Opposite foot off (%)	11.03± 1.85	12.15 ± 3.32	14.6 ± 8.14
Single support (s)	0.40 ± 0.37	0.38 ± 0.06	0.38 ± 0.05
Step length (m)	0.70 ± 0.77	0.72 ± 0.10	0.70 ± 0.68
Step time (s)	0.51 ± 0.48	0.49 ± 0.05	0.48 ± 0.09
Stride length (m)	1.41 ± 0.16	1.42 ± 0.16	1.42 ± 0.15
Stride time (s)	1.02 ± 0.94	1.02 ± 0.11	1.02 ± 0.95
Walking speed (cm/s)	1.39 ± 0.21	1.38 ± 0.21	1.39 ± 0.20

Table 6.7 Temporal spatial parameters under three conditions

An increase in the parameter opposite foot off was noted in the trainer and orthotic group compared to the barefoot group and opposite foot contact decreased in value in condition trainers and orthotic. Step length, foot off, double support and step time increased from barefoot to trainer condition and then decreased in the orthotic group.

Results from the repeated measures ANOVA test showed no significance at all for the data across the three groups. There were also no significance shown between the three groups ($p>0.05$).

Gender differences between the three conditions for temporal spatial data

The table below shows descriptive statistics for the patient barefoot group comparing males and females. An independent *t*-test was conducted to test if there were any significant differences between temporal spatial data between the sexes in each of the three conditions. Only one statistical significance was found in the cadence variable ($p=0.03$), right leg cadence ($p=0.03$) and left leg cadence ($p=0.02$) for the barefoot condition.

Average of right and left	**Gender**	**Mean**	**Std deviation**	**p-value**
Cadence (steps/min)	**Male**	112.57	6.87	0.03
	Female	124.5	11.11	

Table 6.8 Gender statistics for temporal spatial data (barefoot)

No statistical differences existed in the trainers group but cadence showed significance in the orthotics group, right leg ($p=0.03$), left leg ($p=0.04$).

Average of right and left	Gender	Mean	Std deviation	p-value
Cadence (steps/min)	**Male** **Female**	**112.47** **125.42**	**8.78** **11.83**	**0.045**

Table 6.9 Gender statistics for temporal spatial data (orthotics)

Temporal spatial data comparing symptomatic and asymptomatic groups

Foot off (mean normal group= 59.99%, patient group=61.18%, p=0.017) was the only variable to show significance when comparing symptomatic and asymptomatic groups (barefoot condition). Most of the variables were lower in the normal group although cadence was increased in the normal group (119 steps/min, 118 steps/min) compared with the injured group.

6.5. Kinematic data results

Three clinical planes of motion (sagittal, frontal and transverse) were analysed at the pelvis, hip, knee, ankle and foot joints. All joint comparisons were examined bilaterally. The graphs showing wave patterns from the kinematic data of the symptomatic group with barefoot, trainers and orthotic conditions are far too numerous to appear in this chapter. References to a few are throughout the results section but a full concise list of them appears in the appendix.

Comparing asymptomatic and symptomatic subjects with minimum and maximum values.

Comparisons were made with kinematic data from pathological gait cycles against averaged kinematic data from the recently obtained non-pathological gait cycles.

An independent *t*-test was carried out on the two groups comparing the minimum and maximum values of the kinematic data. The results are shown below. The results have been presented by boxplots as this can show any outliers. Outliers are cases with scores that are quite different from the rest of the group, either much higher or lower. In addition to this, boxplots allow us to inspect the pattern of scores within the two groups. It also allows for an indication of the variability in scores within the groups and a visual inspection of the differences between groups using the median level.

Maximum data

The only significant variable in this group was foot progression angle (external angle), asymptomatic (mean = 21.35°, SD 33.01), symptomatic (mean = 5.31°, SD 8.07), t (35.30) = 2.52, p=0.017) equal variances not assumed.

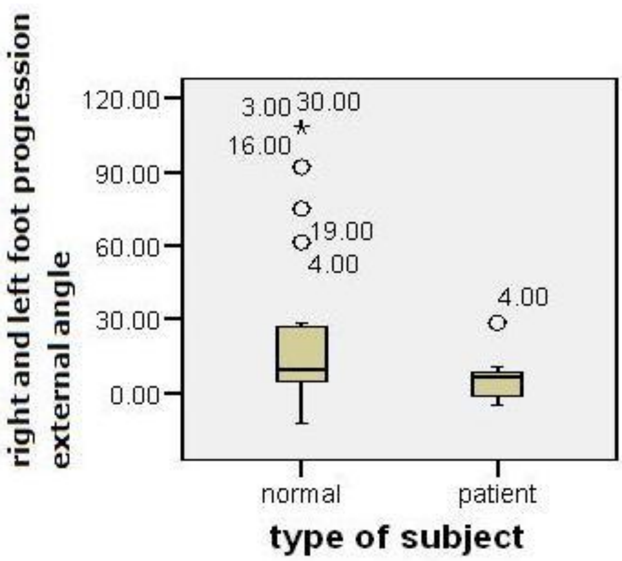


Figure 6.5 Maximum asymptomatic vs symptomatic values of foot progression external angles (°)

Median values for the control group are 9.57° and 6.23 ° for the injured group.

Kinematic graphs showing differences between symptomatic and asymptomatic groups

Samples of these graphs are shown below however a complete list of all graphs can be found in the Appendix.

The *x*-axis of each graph represents the percentage of the gait cycle and the *y*-axis represents the joint angle. Each curve, therefore, shows the variation of the relevant joint angle during the gait cycle. Mean angles are superimposed in each graph. The vertical line in each graph represents the separation between stance and swing phase at approximately 60% of the gait cycle. The red line represents the symptomatic group and the blue line represents the asymptomatic group.

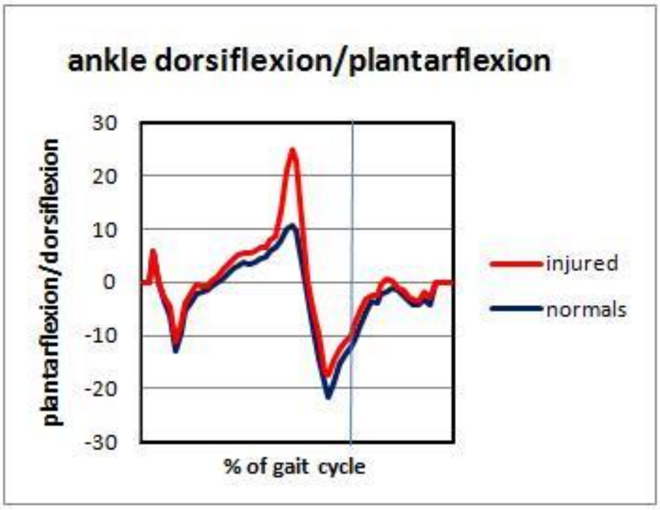


Figure 6.6 Ankle dorsiflexion/plantarflexion (°) in normal and injured groups

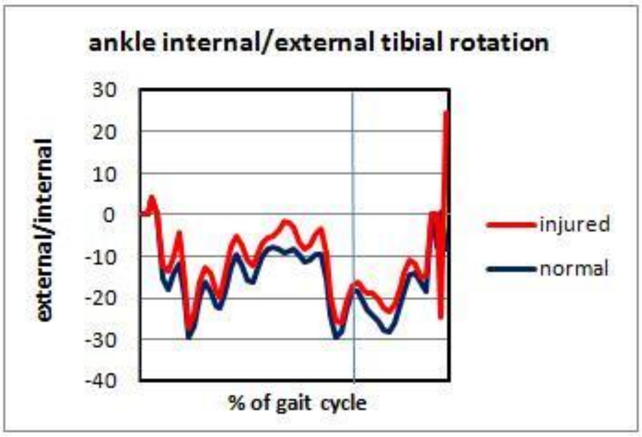


Figure 6.7 Ankle internal/external tibial rotation (°) in normal and injured groups

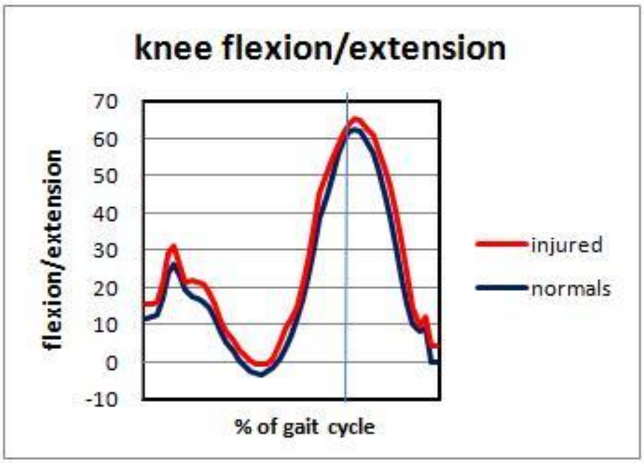


Figure 6.8 Knee flexion/extension (°) in normal and injured groups

Figures 6.6, 6.7 and 6.8 display the results from the asymptomatic group compared with the results from the symptomatic group. Maximum ankle dorsiflexion is much higher in the injured group but maximum ankle plantarflexion is lower during stance phase. Tibial rotation (figure 6.7.) shows the injured group with increased maximum values throughout the stance phase and the non-injured group displaying higher minimum values. Figure 6.8 displays knee flexion angles and again show higher maximum values throughout the stance phase in the injured group.

Comparison of maximum values between male and female symptomatic subjects

Independent *t*-tests were performed to compare differences in kinematics in male and female symptomatic subjects. In the maximum group, five of the variables examined had significant differences between the sexes.

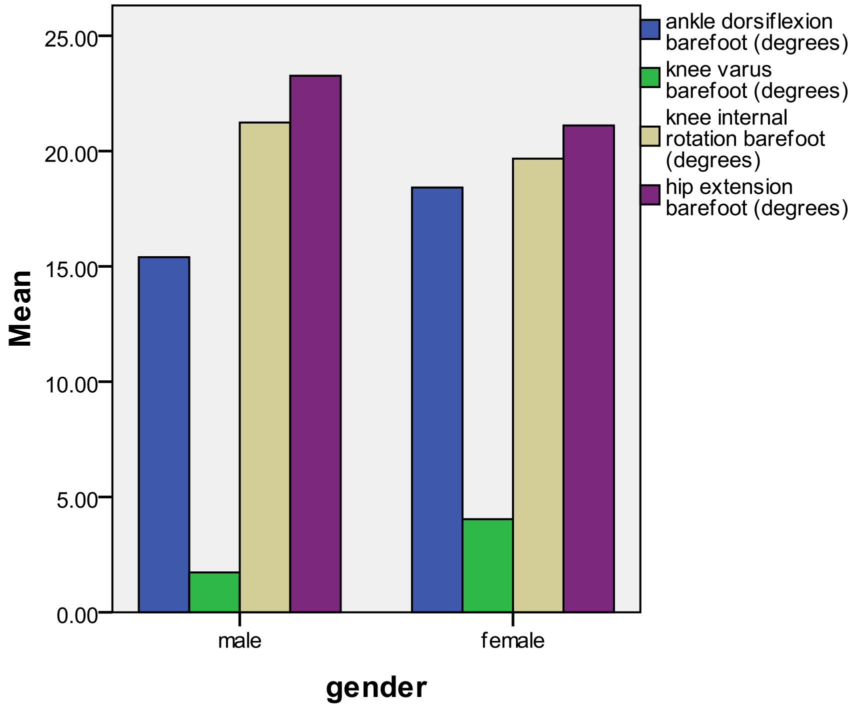


Figure 6.9 Graph showing significant variables between males and females in barefoot walking

Variable barefoot	Male (°)	Female (°)	*P* value
Ankle dorsiflexion	6.9	28.09	0.010
Knee varus	4.21	1.20	0.050
Knee internal rotation	23.42	17.17	0.009
Hip extension	37.5	43.37	0.022

Table 6.10 Maximum variables significantly different between males and females walking barefoot

From the table above it was shown that females were significantly greater than males in ankle dorsiflexion and hip extension and males were significantly greater in knee varus and knee internal rotation.

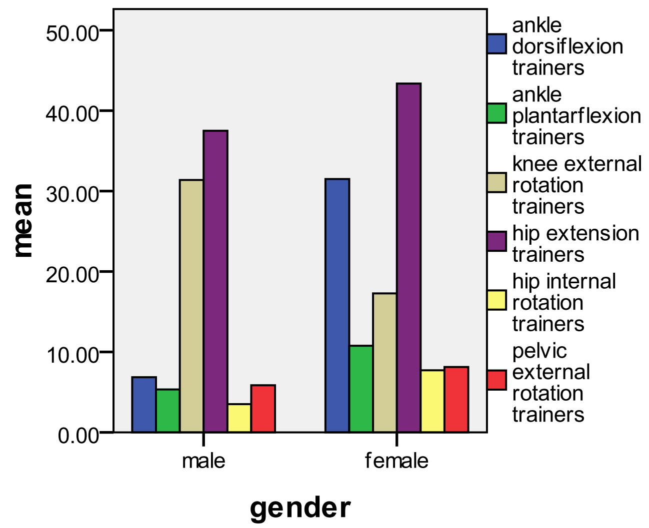


Figure 6.10 Graph showing significant variables between males and females in walking with trainers (°)

Variable trainers	Male (°)	Female (°)	*P* value
Ankle dorsiflexion	6.86	31.5	0.016
Ankle plantarflexion	5.34	10.77	0.008
Knee external rotation	31.38	17.28	0.003
Hip extension	38.63	44.10	0.021
Hip internal rotation	3.51	7.72	0.007
Pelvic external rotation	5.86	8.13	0.030

Table 6.11 Maximum variables significantly different between males and females walking with trainers

From Table 6.11 females exhibited greater values for all variables except knee external rotation.

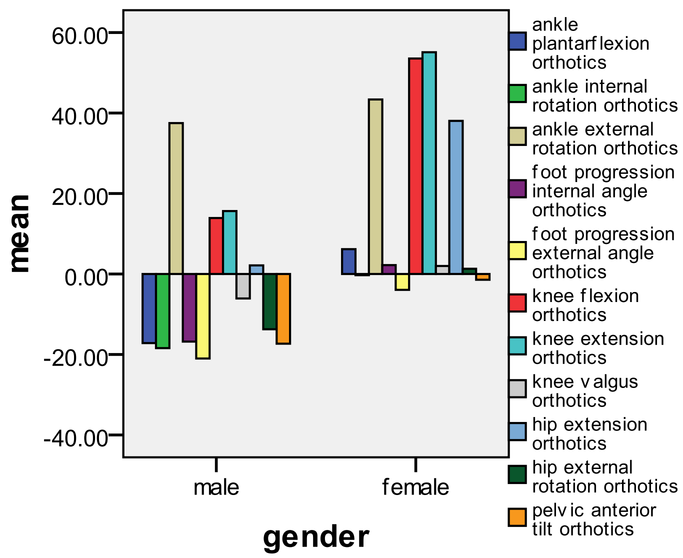


Figure 6.11 Graph showing significant variables between males and females in walking with trainers and orthotics (°)

In Table 6.12 male angles are far greater than female angles with the exception of knee flexion and extension.

Variable orthotics	**Male (°)**	**Female (°)**	***P* value**
Ankle plantarflexion	-17.17	6.17	0.006
Ankle abduction	-5.03	0.47	0.024
Ankle internal rotation	-18.42	-0.29	0.008
Ankle external rotation	-25.05	-5.62	0.006
Foot progression internal angle	-16.8	2.21	0.002
Foot progression external angle	-21.00	-3.94	0.010
Knee flexion	13.91	53.55	0.014
knee extension	15.64	55.11	0.012
Knee valgus	-6.08	1.99	0.035
Hip extension	2.14	38.05	0.004
Hip external rotation	-13.73	1.3	0.031
Pelvic anterior tilt	-17.32	-1.43	0.032

Table 6.12 Maximum variables significantly different between males and females walking with trainers and orthotics

Analysis of kinematic data

Maximum internal tibial rotation was progressively decreased between the barefoot and orthotic conditions in 11 out of the 15 subjects (73%) and maximum eversion values decreased but in only 3 out of 15 (20%) subjects in the orthotics condition. However, despite the small number of reduced eversion values, 7 out of the sample coupled internal tibial rotation with eversion.

In a few subjects, knee external rotation, hip extension and pelvic obliquity up, values were increased in the trainer condition but decreased in the orthotic condition although were still a higher value than in the original barefoot condition (see table on p210).

MAXIMUM VALUES OF KINEMATIC DATA

When testing ankle plantarflexion, Mauchlys test indicated that the assumption of sphericity had been violated ($X(2) = 16.27$, $p=0.001$) therefore the degrees of freedom were corrected using Greenhouse- Geisser estimates of sphericity ($e=0.58$). The reason the GG test was chosen was that $e<0.75$ (Field, 2000). The results show that there was a significant effect on the three conditions on ankle plantarflexion, $F(1.17, 16.34) = 12.98$, $p=0.002$. The maximum ankle plantarflexion angle was significantly increased between groups barefoot and orthotics ($p= 0.006$) and trainers and orthotics ($p= 0.009$).

As shown in Figure 6.12, subject 9 is an outlier, so the same test was repeated without the data from subject and found there was still a significant difference between the three conditions on ankle plantarflexion , F(1.15, 15.03)= 13.24 p=0.002.

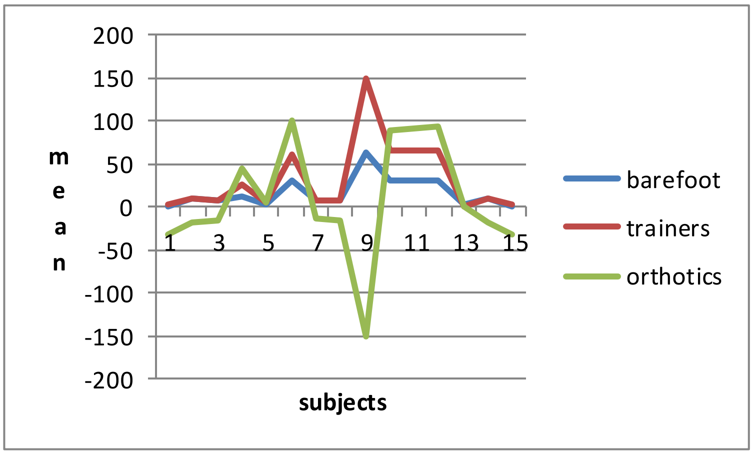


Figure 6.12 Mean data for maximum ankle dorsiflexion/plantarflexion angles (°) of three walking conditions

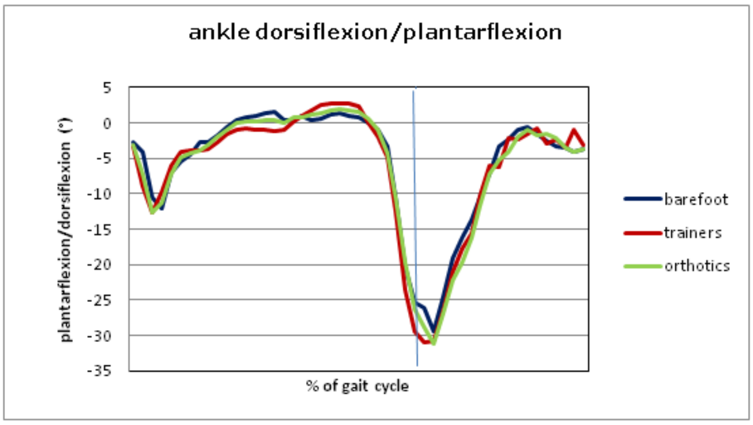


Figure 6.13 Ankle dorsiflexion/plantarflexion over three conditions

No significant differences were found for maximum ankle dorsiflexion (p<0.05).

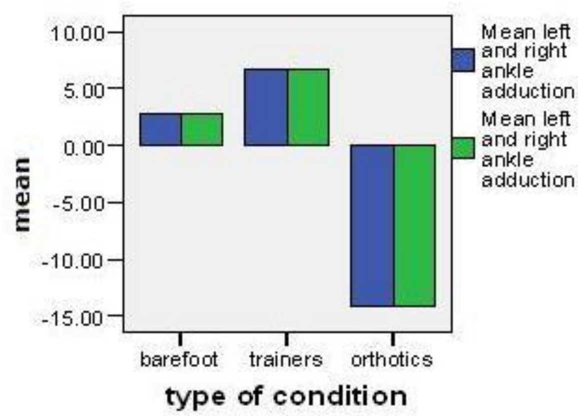


Figure 6.14 Maximum ankle abduction/adduction over three conditions

Mauchlys test was not violated so sphericity was assumed. The results show that there was a significant difference between the groups for abduction F (2, 28) = 9.01, p=0.001). It was also shown that there was a significant difference of maximum ankle abduction between the groups of barefoot to orthotics (p=0.012) and trainers to orthotics (p= 0.017). There was no significance for maximum adduction of the ankle although angle values decreased under all three conditions, barefoot to trainers (p=0.04), barefoot compared to orthotics (p=0.0020 and trainers compared to orthotics (p=0.023).

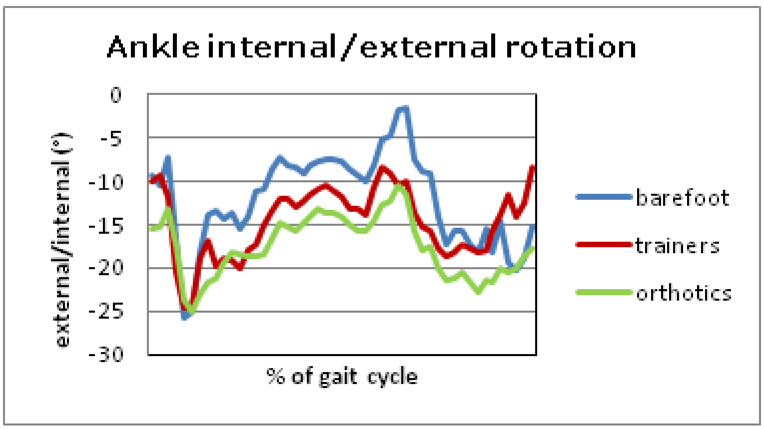


Figure 6.15 Ankle internal/external rotation over three conditions

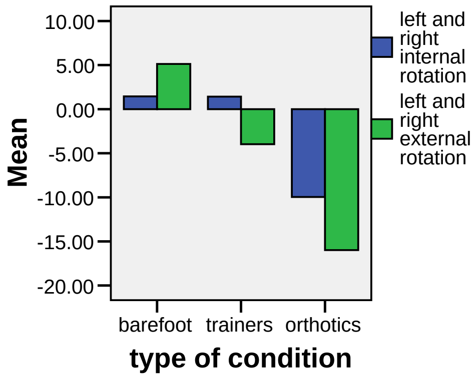


Figure 6.16 Mean data for maximum ankle internal/external angles of three conditions

Mauchlys test was not violated for ankle internal rotation so sphericity was assumed. The results show that there was significant difference between the groups $F(2, 28) = 8.32$, $p=0.001$). It can be seen that there was significant decrease of internal rotation between the groups of trainers to orthotics ($p= 0.007$) and barefoot to orthotics ($p= 0.012$). Ankle external rotation was also statistically significant $F(1.4, 19.6) = 13.67$, $p= 0.001$. The maximum external rotation angle was significantly increased under the barefoot-orthotics ($p=0.003$), barefoot to trainers ($p=0.03$) and trainers to orthotics ($p= 0.014$) conditions.

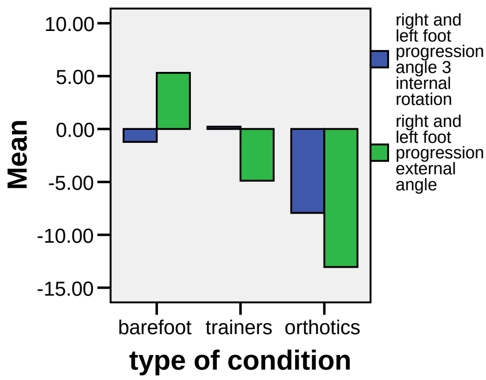


Figure 6.18 Mean data for maximum foot progression (°) over three conditions

The graphs above, foot progression angles were statistically all significant F(1.5, 20.29)=10.36, p=0.002; F(2, 28)=12.16, p=0.001). The maximum internal foot progression angle was significanctly increased under groups barefoot to orthotics (p=0.04) and trainers to orthotics (p=0.01). The maximum external foot progression angle increased significantly between groups barefoot to trainers (p=0.01) and barefoot to orthotics (p=0.004).

Knee flexion F (1.02, 14.32) = 12.84, p= 0.03 and knee extension F (2, 28) = 11.11, p= 0.001 show both have significance between the three groups. Peak knee flexion angles were significantly decreased between barefoot and orthotics (p=0.01) and trainers and orthotics (p=0.006) and knee extension angles were significantly increased between barefoot and trainers (p= 0.002) and reduced between trainers to orthotics (p=0.09).

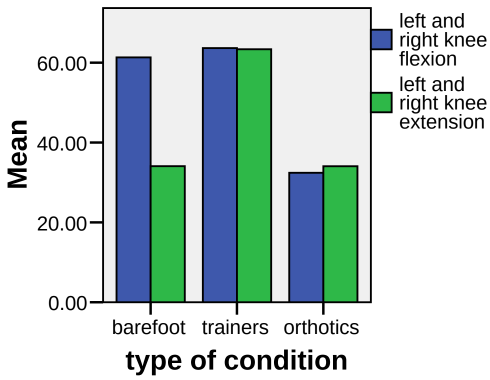


Figure 6.19 Mean data for maximum knee flexion/extension angles (°) of three conditions

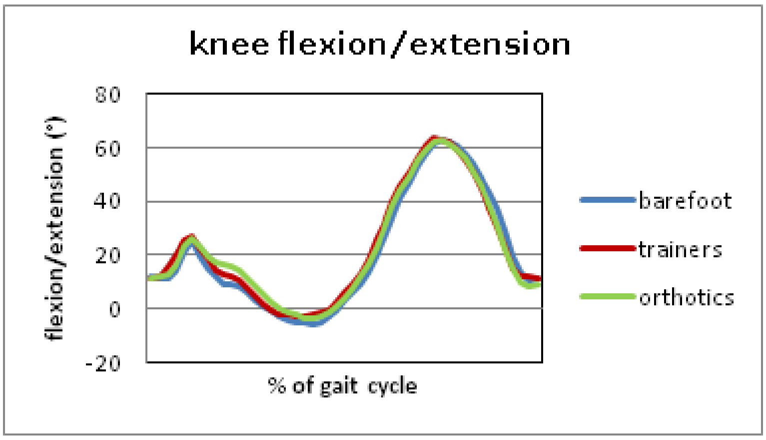


Figure 6.20 Knee flexion/extension over three conditions

The maximum knee varus angle was significantly increased F (2, 28) = 5.99, p= 0.007 under trainers to orthotics (p =0.03) conditions. Knee valgus displayed no statistical significance (p>0.05).

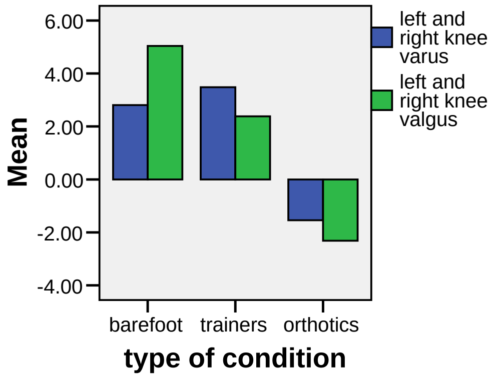


Figure 6.21 Mean data for maximum knee varus/valgus angles (°) over three conditions

The peak internal rotation angle shows significantly reduced values under the barefoot to orthotics (p= 0.023) conditions and trainers to orthotics (p= 0.002) conditions.

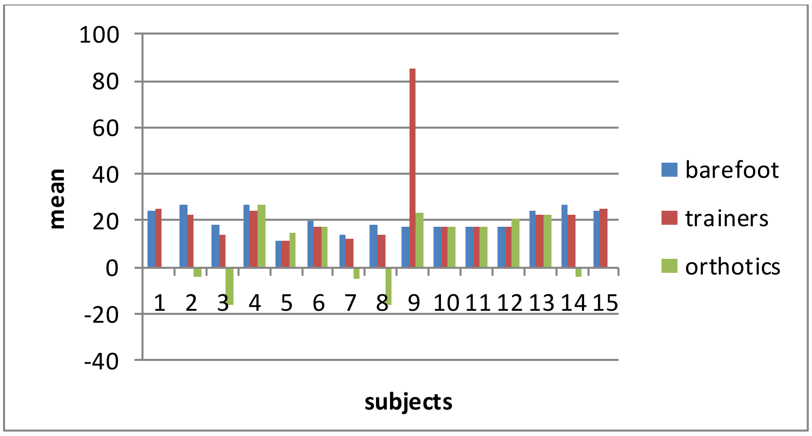


Figure 6.22 Maximum values of internal/external knee rotation (°) angles over three conditions

The graph clearly shows an unusual peak of rotation in the trainer group from subject 9 (85.3°). The data was run again with ANOVA without subject 9 and although different, still statistically significant $F(1.02, 13.23) = 10.75$, $p=0.006$). There were also changes within the group conditions data- barefoot compared to orthotics ($p= 0.001$), barefoot compared to trainers ($p=0.05$) and trainers compared to orthotics ($p=0.023$).

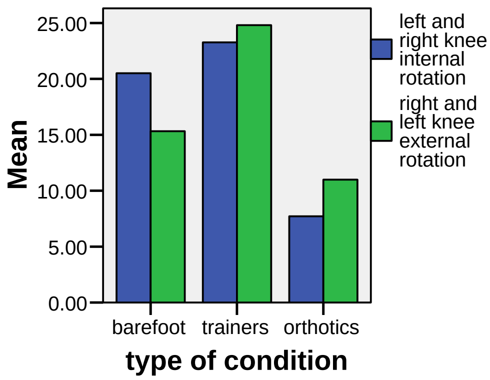


Figure 6.23 Mean data for maximum knee internal rotation angles (°) over three conditions $F(2, 28) = 6.60$, $p=0.004$

Hip flexion values were not statistically significant ($p=0.151$).

Peak hip extension values were significantly increased under conditions barefoot to orthotics (p=0.012), reduced under trainers to orthotics (p=0.010) and increased under barefoot to trainers (p= 0.002); F (1.00, 14.02) = 12.13, p=0.004).

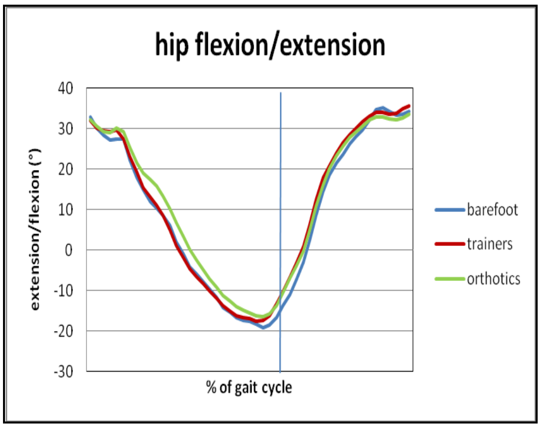


Figure 6.24 Hip flexion/extension angle over three conditions

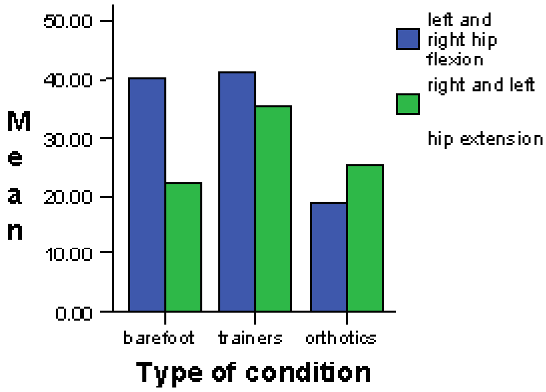


Figure 6.25 Mean data for maximum hip flexion/extension angles (°) over three conditions

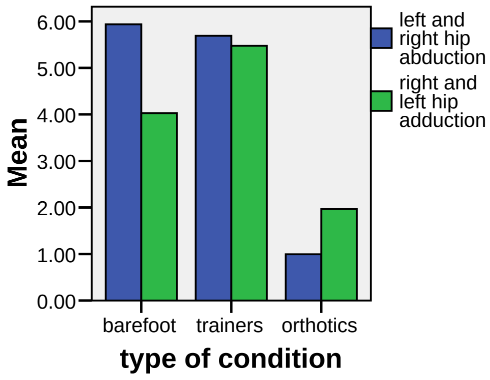


Figure 6.26 Mean data for maximum hip abduction angles (°) over three conditions

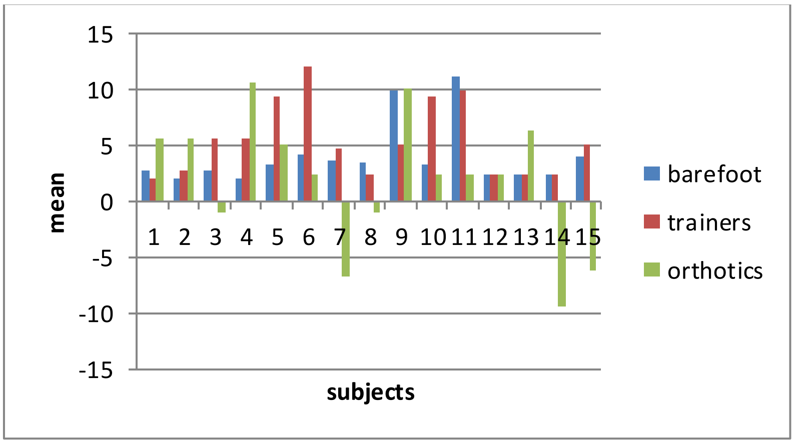


Figure 6.27 Maximum hip adduction/abduction angles (°) over three conditions

Hip abduction did not show any statistical significance under the three different conditions. Hip adduction was significant $F (1.17, 16.38) = 9.74$, $p=(0.005)$. Peak hip adduction decreased significantly under conditions barefoot to orthotics ($p=0.02$) and decreased again with trainers compared to orthotics ($p=0.014$).

Maximum hip internal rotation was not significant however, maximum hip external rotation was $p = (0.002)$.

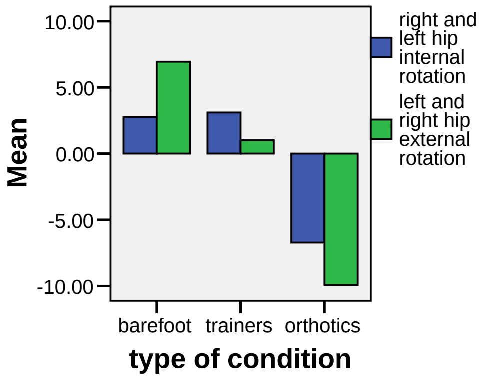


Figure 6.28 Mean data for maximum hip internal/external rotation (°) over three conditions

Pelvic anterior tilt, F (1.37, 19.24) = 12.43, p= 0.001 and pelvic obliquity up, F (2, 28) = 9.52, p=0.001 both displayed statistical significance. Both had the same pairwise comparisons, barefoot to trainers (p= 0.28), (p=0.23) and barefoot to orthotics (p=0.002), (p=0.006).

Pelvic posterior tilt, pelvic internal or external rotation did not show any significance between the three conditions.

6.6. Extra marker set on foot and leg

The extra marker placed on the top of the navicular was able to measure the height of the navicular from the floor in millimetres. Tests were run on static measures and dynamic conditions. The static measurements were available from the calibration trial therefore were only available in the barefoot condition. Paired *t*-tests were conducted to test for any differences between pathological sides and unaffected sides.

In the static scores, no significance was shown between the two sides (p=0.84). Minimum and maximum values were very similar in right and left sides (left, min 8.60mm, max 21.10mm, mean 14.6mm; right, min 9.19mm, max 21.9mm, mean 14.9mm).

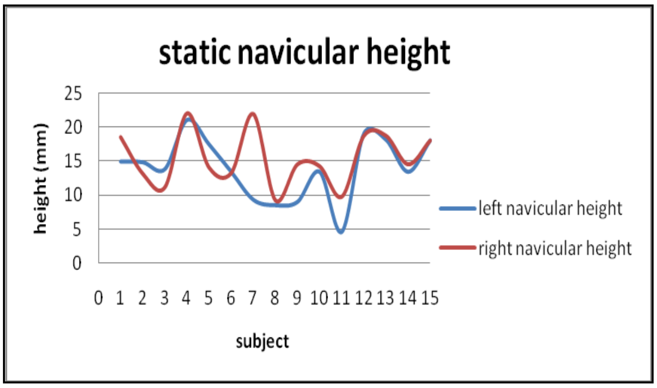


Figure 6.29 Static navicular height values in 15 subjects

An independent t-test was carried out to test for differences in gender and although male height was higher than female height in both injured and non-injured sides, no significant differences were shown (p>0.05).

Values were sought from the maximum and minimum groups in the dynamic trails over the three experimental conditions. In the maximum and minimum groups both found statistically significant differences. The maximum group showed significance in the trainer condition between right and left (p=0.045) and the minimum group for barefoot condition between right and left (p=0.013)

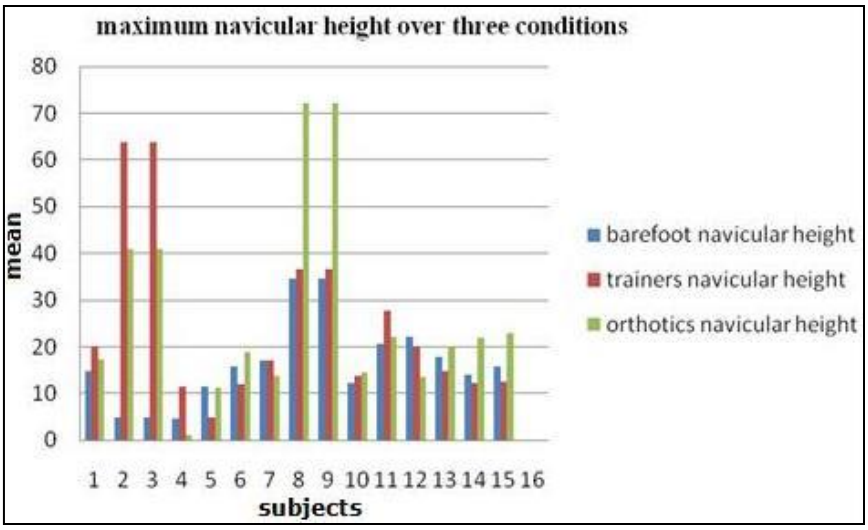


Figure 6.30 Peak navicular height (cm) values over three conditions

The graph above shows that 60% of subject navicular height values reduced when in the trainer condition compared to the barefoot condition. This could be due to the marker on the shoe not being placed in the same position as there was no cut out for the navicular bone on the trainer. 55% of values increased when wearing orthotics compared to the barefoot condition which is what would be expected and 73% of measurements increased between the trainer and the orthotic condition. It was expected that the height would increase when in the trainer and orthotic condition.

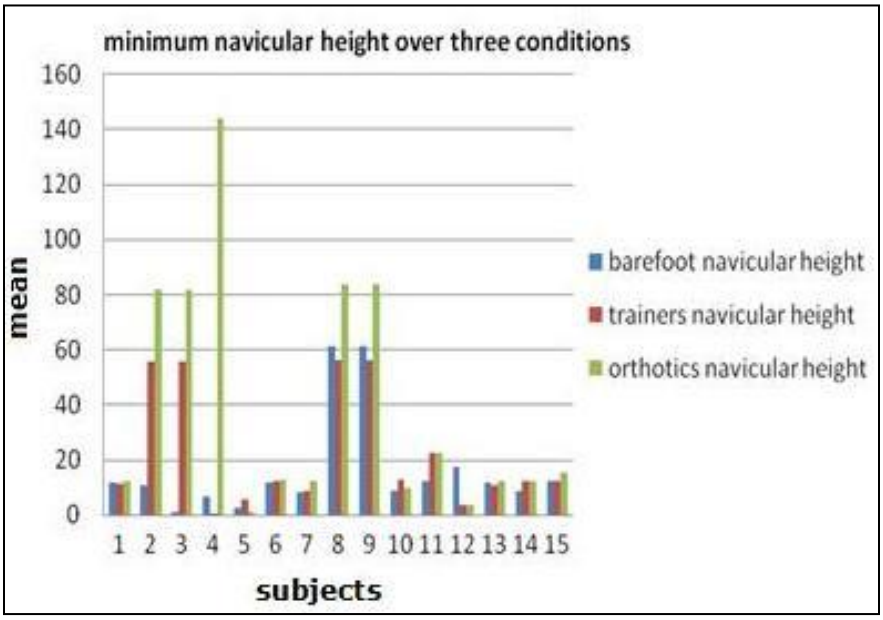


Figure 6.31 Minimum navicular height values (cm) over three conditions

Repeated measures ANOVA was carried out to detect any comparisons between the three groups but no comparison were found in either of the maximum or minimum groups. There were differences between the means of the three experimental groups but the standard error was quite high.

Calcaneal Inversion and Eversion results

A matched paired *t*-test was conducted to test for differences between the pathological leg and the unaffected leg in which it was shown no significances in either the maximum and minimum values (max group, barefoot R & L ($p=0.09$); trainer R & L, ($p=0.57$); orthotic R & L, ($p=0.92$); min group, barefoot R & L, ($p=0.55$); trainer R & L, ($p=0.26$); orthotic R & L, ($p=0.75$). A repeated measures ANOVA was also performed to see if there were any significant differences between each condition of barefoot, trainers and trainers with orthoses. The results were as shown below.

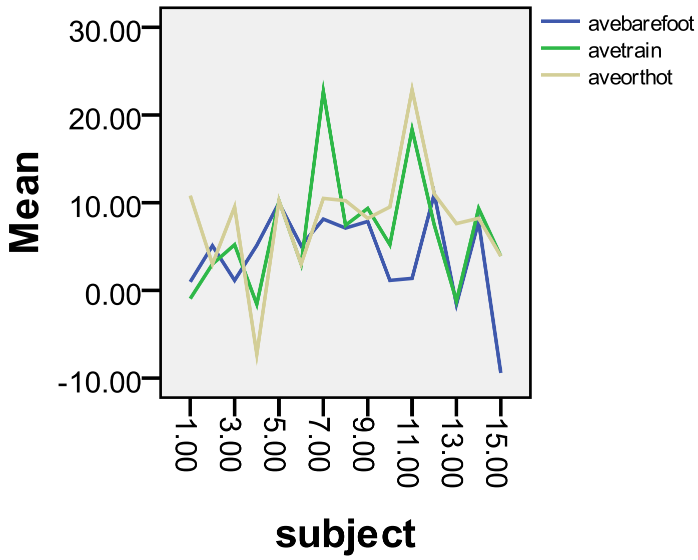


Figure 6.32 Maximum mean values of inversion/eversion (°) over three conditions

Condition	N	Min (°)	Max (°)	Mean (°)	Std. Dev
Barefoot	15	-9.42	11.00	4.0	5.2
Trainers	15	-1.58	22.70	6.7	6.8
Orthotics	15	-7.30	22.90	8.0	6.3
Valid N (listwise)	15				

Table 6.13 Maximum values for inversion/eversion over three conditions

The graph clearly shows a similarity between the trainer and the orthotic groups however no significance was shown between any of the groups ($F = (2, 28) = 2.62$, $p=0.09$). The descriptive statistics show that the mean values increase throughout the three conditions, something that was not expected.

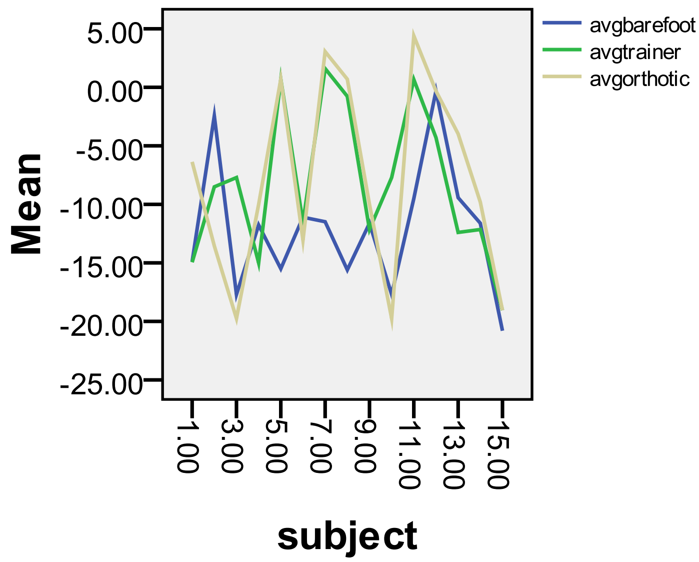


Figure 6.33 Minimum mean values of inversion/eversion (°) over three conditions

Condition	N	Min (°)	Max (°)	Mean (°)	Std. Dev.
Barefoot	15	-20.80	-.32	-12.0	5.4
Trainer	15	-19.05	1.60	-8.2	6.5
Orthotic	15	-19.75	4.40	-7.7	8.3
Valid N (listwise)	15				

Table 6.14 Minimum values for inversion/eversion over three conditions

The mean values for the minimum group reduced throughout the three conditions with eversion values reducing from -12.1° to -7. 79°.

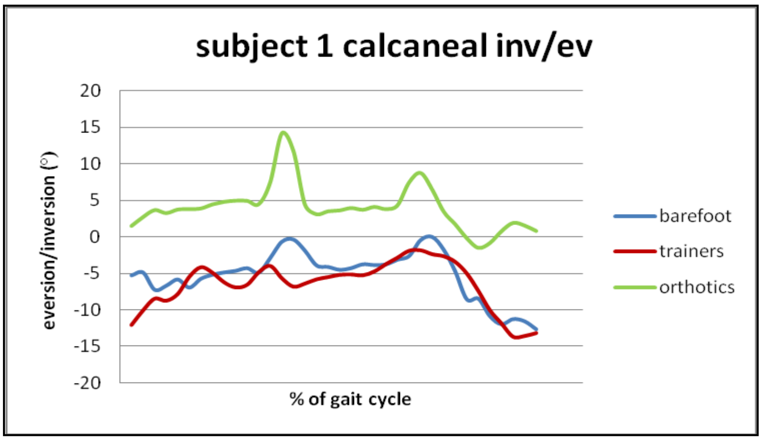


Figure 6.34 Inversion/eversion angles of one random subject

Figure 6.34 shows maximum inversion/eversion values throughout 100% of the stance phase for one random subjects. Negative values denote eversion. In the frontal plane, calcaneal inversion and eversion displayed differences between barefoot, trainers and orthotics. Looking at the subject 1 graph (Figure 6.34.), the barefoot condition started at heel strike in an everted position and the curve inverted and everted through to mid stance, started to evert again, and push off in an everted position. In the orthotic condition, the rearfoot lands in an inverted position and immediately started inverting and peaking at mid-stance, everting through to late stance and push off. The trainers pattern follows a very similar trend as the barefoot pattern except that it started in a more everted position and didn't peak as high as the barefoot condition. Both barefoot and trainers start in an everted position but the orthotic condition starts heel strike in a very inverted position. The rest of the graph is very similar except for mid- to late stance where trainers and orthotics are in an inverted positon but the barefoot is everting.

The final two additional markers on the back of the calf measured the calf to calcaneus angle sometimes referred to as the rearfoot angle. This was the 2D angle (see chapter 4 section 4.6). This explains what the rearfoot angle is and why it was being measured.

A matched paired *t*-test was conducted to test for differences between the pathological leg and the unaffected leg in which it was shown no significances in the maximum condition but in the minimum group of values, right and left trainers showed significant differences (mean right 8.33°, left 6.46°, p=0.034).

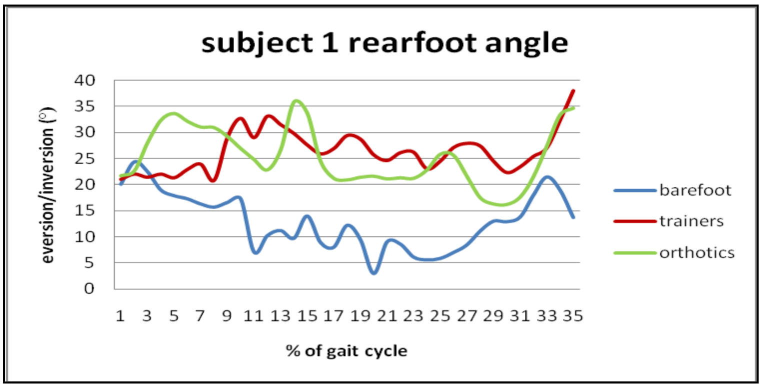


Figure 6.35 Rearfoot angle of one random subject over three walking conditions

Figure 6.35 shows subject 1 rearfoot angle throughout the stance phase over three conditions and figure 6.36 displays the mean of the rearfoot angle of 15 subjects.

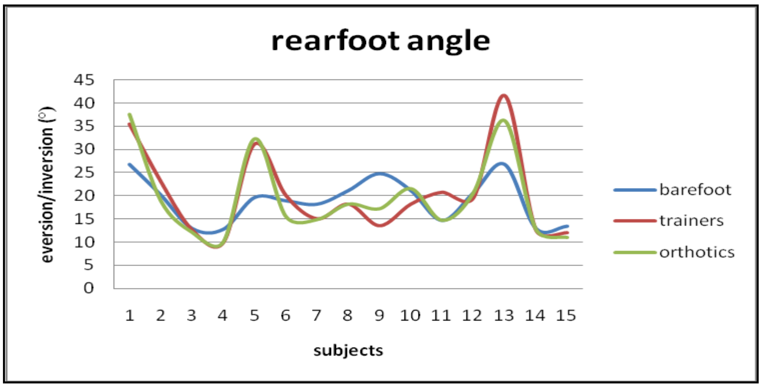


Figure 6.36 Mean values of rearfoot angle of 15 subjects over three walking conditions

A repeated measures ANOVA was also performed to see if there were any significant differences between each condition of barefoot, trainers and trainers with orthoses. The results were as shown below.

Maximum values

Condition	N	Minimum (°)	Maximum (°)	Mean (°)	Std. Deviation
barefoot	15	12.72	26.75	18.9	4.8
trainer	15	9.94	41.65	20.3	9.2
orthotic	15	9.94	37.55	19.5	8.8
Valid N (listwise)	15				

Table 6.15 Maximum values of rearfoot angles over three conditions

Mean maximum values increase in inversion values from barefoot to trainers and decrease slightly when wearing orthotics but not enough to be significant between the three conditions (p=0.552).

Minimum values

Condition	N	Minimum (°)	Maximum (°)	Mean (°)	Std. Deviation
Barefoot	15	1.54	7.68	5.0	2.4
Trainer	15	1.16	19.90	7.3	6.0
Orthotic	15	.36	14.45	6.0	4.0
Valid N (listwise)	15				

Table 6.16 Minimum values of rearfoot angles over three conditions

Minimum values had no significant differences (p=0.144). The maximum value for the trainer and orthotic group were both much higher than the barefoot group. The maximum rearfoot angle is much more inverted when wearing trainers compared to barefoot however the angle starts to evert when wearing the orthotic.

Due to the large amounts of results shown, a concise summary of the repeated measures ANOVA comparing kinematic data over three experimental conditions, barefoot, trainers and orthotics is provided.

Maximum values

	Variable	Maximum (p value)	conclusion	Sig value of comparisons (p)
ankle	Dorsiflexion	No sig		
	Plantarflexion	0.02	↑1-3 ↓ 2-3	0.006, 0.009
	Adduction	No sig		
	Abduction	0.08	↑1-3 ↑2-3	0.012, 0.017
	Int rot	0.001	↓2-3 ↓1-3	0.007, 0.012
	Ext rot	0.001	↑1-3 ↑1-2↑2-3	0.003, 0.03, 0.014
Foot prog	prog int	0.002	↑1-3 ↑ 2-3	0.04, 0.01
	prog ext	0.001	↑1-2 ↑1-3	0.01, 0.004
Knee	Flexion	0.003	↓1-3 ↓2-3	0.01, 0.006
	Extension	0.001	↑1-2	0.002, 0.09
	Varus	0.007	↑2-3	0.03
	Valgus	0.001	↓1-3 ↓2-3	0.017, 0.02
	Int rot	0.004	↓1-3 ↓2-3	0.023, 0.002
	Ext rot	0.002	↑1-2 ↓2-3	0.02, 0.019
Hip	Flexion	No sig	↑1-2 ↓1-3	0.024
	Extension	0.004	↑1-3 ↑1-2↓2-3	0.012, 0.002,0.01
	Abduction	No sig	↑1-2	0.04
	Adduction	0.005	↓1-3 ↓2-3	0.02, 0.014
	Int rot	Not sig	↓1-3	
	Ext rot	0.02	No comparisons	

Pelvis	Ant tilt		↑1-2 ↑1-3	0.02, 0.002
	Post tilt	No sig		
	Obliq up	0.001	↑1-2 ↑1-3	0.03, 0.006
	Obliq down	No sig		
	Int rot	No sig		
	Ext rot	No sig		

↑= angle has increased 1-barefoot, 2-trainers, 3-trainers with orthotics

↓= angle has decreased

6.7. Discussion

The primary purpose of this study was to evaluate the biomechanical effects, if any, of wearing orthoses and trainers compared to walking barefoot. This was achieved by analysing the 3D kinematics and the temporal parameters for any changes during the three experimental conditions. This may then provide a more comprehensive explanation in the successful treatment of AKP.

The other main aim was to analyse the new definitive foot model measuring navicular height, rearfoot angle and inversion and eversion in the frontal plane.

6.7.1. Clinical assessment for pathological group

In the clinical assessment, it was noted that in the symptomatic group there were significant differences between right and left sides during Q-angle (w/b), rearfoot angle measurement and hip flexion compared to no significant differences in the asymptomatic group. It was expected that there might be some differentiation in the patient group because these subjects all had a concurrent knee injury and both sides may not be the same. This could be due to the fact that it was established in the literature review that due to altered biomechanics, vastus lateralis often alters the pull of the Q-angle which could therefore create a larger Q-angle value.

Influence of gender on symptomatic group

Ball & Johnson (1996) demonstrated a small but significant gender difference with females exhibiting a greater range of motion than males in all the age groups. It was found in this study that only five of the variables had a significant difference between the sexes.

When comparing gender on the symptomatic groups, the same variables as in the normal group were significant, ankle width, q-angle w/b and supine, leg length and navicular height.

Navicular height was significantly higher in the male group compared with the female group. Nielson et al (2009) agreed with this as they stated that as foot length increases, navicular height increases. This was expected as discussed in the asymptomatic group. Zeybek et al (2008) carried out a study on 249 subjects measuring right and left navicular height. They found significant differences between their measurements between men and women. Their static navicular height values were not consistent with this study's results as they had much higher scores but their measurement was non-weight-bearing. Perhaps a small methodological flaw in the present study was that body weight was not taken into account when measuring weight-bearing navicular height therefore it was more than likely that body weight alone reduced the height of the navicular.

It was of course expected that women's Q-angle would also be significantly different in this group. Leg length and ankle width which has been discussed earlier are dependent on height (r= 0.95 leg length-height and r= 0.75 ankle width- height) so again it was expected to see a difference between the genders within these variables.

Influence of gender on symptomatic compared with asymptomatic group

Comparing the male and female values within both groups, although not significant, the females exhibit higher Q-angle values. The female values are consistent with other studies (La Brier & O'Neill 1993) and (Hughston et al 1984) state that anything more than 15 in women is abnormal. However, the male value of Q- angle weight-bearing was lower in the AKP group than the normal group and the females had a higher value in the AKP group than the normal group. The Manual of Orthopaedic Surgery (1972) considers angles greater than 15 degrees in men and 20 degrees in women pathological however, that does appear to be a little inaccurate (Horton & Hall 1989).

Gender		Q angle weight bearing right and left (°)	Q angle supine right and left (°)
Male	Mean	9.75	9.93
	N	8	8
	Std. Deviation	2.32	2.24
Female	Mean	16.92	15.78
	N	7	7
	Std. Deviation	3.95	3.37
Total	Mean	13.10	12.66
	N	15	15
	Std. Deviation	4.81	4.064

Table 6.17 Q-angle weight- bearing and supine for patient group

Theoretically, a higher Q-angle increases the lateral pull of the quadriceps femoris muscle on the patella and potentiates patellofemoral disorders (Horton & Hall 1989).

There was a 1.6mm difference in navicular height between the asymptomatic group and the symptomatic group. It was expected for the asymptomatic group to have a higher value as the patient group consisted of mostly over pronators, which would result in a smaller navicular height score.

Differences in clinical assessment between symptomatic and asymptomatic groups

Rearfoot angle, ankle plantarflexion and 1st MTP joint dorsiflexion were all significant variables when comparing both symptomatic and asymptomatic groups.

Rearfoot angle in the injured group was 2.2° everted higher than the non-injured group, ankle plantarflexion and 1^{st} MTP joint dorsiflexion was increased in the non-injured group compared to the injured group. Despite the frequent clinical use of the rearfoot angle, its direct relationship to abnormal rearfoot motion during locomotion and its relationship to lower extremity injury are sparse (Cornwall & McPoil 2004).

Minimum and maximum rearfoot angle values (inversion/eversion) are shown in table below. (Negative values indicate inversion and positive values indicate eversion).

Name of study	N	Min (°)	Max(°)	Mean(°)
Present study	n=30 (normal) n=15 (injured)	-2.5 4.5	15 15	6.85 9.06
Chuter (2010)	n=20(normal) n=20 (pronated)	3 7	7 14	4.95 10.71
McPoil & Cornwall (2004)	n= 82	-0.2	11.8	6.3
Sobel (1999)	n=88	-1	14	6.07

Table 6.18 Maximum, minimum and mean rearfoot angles

The results of the present study state that in the normal group the maximum value is 15 degrees, which is extremely high, compared to the other studies shown in the table. However, this particular value is an outlier when if it was removed, the maximum value for rearfoot was 11 degrees of eversion. Root et al (1977) proposed that the angle of the rearfoot in quiet standing should be in the range of 0-2 degrees either inverted or everted. This appears to be a very low and unrealistic value and is not consistent with other empirical studies. Sobel et al (1999) reported very different values to Root and colleagues and found maximum values to be 14 degrees everted. The present study's results compare favourably with the shown rearfoot angle values.

Ankle plantarflexion was reported to be reduced significantly in the patient group by approximately four degrees. Lun et al (2004) reported similar reductions in plantarflexion ROM between the injured and non-injured subject.

Caylor et al (1993) found no significant difference in Q-angle between asymptomatic subjects (11.1 +/- 5.5 degrees) and symptomatic subjects (12.4 +/- 5.1 degrees) ($p = 0.07$). They concluded that increased Q-angles were not responsible for AKP in this group of patients. Other factors were hypothesized to be responsible for their symptoms. The results of this present study agreed and found that no statistical differences were found between either groups. The female symptomatic group had higher values for both supine and weight bearing methods and the male symptomatic group also had a higher weight bearing value but a lower supine value.

Gender differences in temporal-spatial data

Because of there being distinct gender differences in skeletal dimensions, the walking patterns of the two genders are likely to differ (Cho et al 2004). In all three conditions of barefoot, trainers and orthoses, cadence was a constant statistical significant variable. The female group exhibited many more steps per minute (124 compared with 112 barefoot; 125 compared with 112 trainers and 124 compared with 112 orthotic group). Murray et al (1964) and Murray et al (1970) studied data that showed that females walked slower with shorter step lengths but with greater cadence than males. This however was proven without any statistical analysis but Oberg et al (1993) found consistent results as Murray and colleagues. In the trainer condition, stride time and step time were deemed significant and the females were slower in both variables than the males. In the orthotic condition, stride time was significant with females slower than males and opposite foot off and single support also significant.

The results show that no significant differences existed between the three conditions for temporal spatial data in the pathological group. Chen et al (2010) found that stride length was significantly longer in the shoe condition and the insole condition than the equivalent results in the barefoot group (insoles compared to barefoot ($p= 0.02$) and shoes compared to barefoot ($p= 0.03$). The shoes however were custom made for each subject and the insoles had been moulded for each subject. An interesting observation in this study was that there was a 4% reduction in step frequency, approximately 3% increase in step length and a 1cm increase in stride length between barefoot and trainers. Majumdar et al (2006) explained that this increase in step and stride length may be due to added cushioning, snugness of fit and comfort of the shoe. They tested military boots against barefoot but the same snugness and comfort could be applied to the fit of a trainer.

When comparing the symptomatic group with the asymptomatic group, it was also found that foot off was the only variable to increase significantly. The results show that temporal spatial data was not particularly affected by the injured subjects and their pathological legs.

6.7.2. Kinematics of symptomatic subjects

Gender differences

Ferber et al (2003) also looked at gender differences between normal kinematic data in running and found significant maximum values in hip adduction, knee abduction and hip internal rotation. They hypothesised that females would have a lower internal knee rotation angle due to the greater femoral internal rotation. In the present study, the results were consistent with the hypothesis (Mean hip internal rotation – male 2.90 degrees female 5.31 degrees, mean knee internal rotation – male 23.42 degrees female 17.17 degrees) although it must be noted that the present study looked at symptomatic subjects walking and Ferber et al (2003) studied a normal group running. Kerrigan et al (1998) reported that females had significantly higher hip flexion and lower knee extension. This study reported a significantly higher hip flexion and although not significant, a higher knee extension in women.

In the orthotic condition, knee flexion was very low in the male group and knee extension was very high in the female group. This agrees with Malinzak (2001) who reported less peak knee flexion in females.

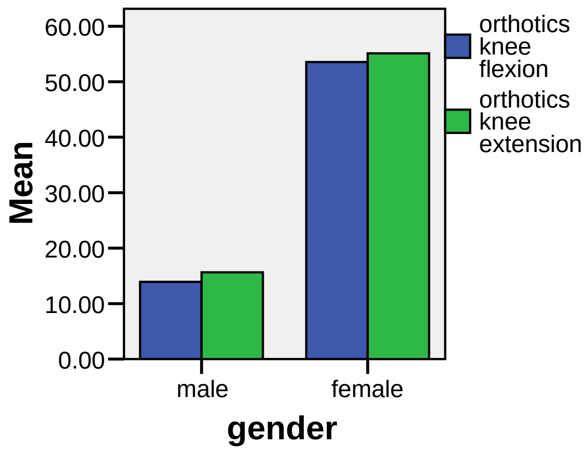


Figure 6.38 Significant mean differences between males and females in orthotic condition

In the symptomatic barefoot group, knee flexion values were slightly higher for both sexes with 61.56° for males and 61.04° for females. Knee extension values were 21.31° for males and 48.68° for females. In the normal group, knee peak flexion and extension values were for males 60.29°, 17.07°, and females 53.88° and 24.90° respectively. In the trainers group, knee flexion values were slightly raised from the barefoot group and knee extension values were raised for both sexes but especially in the male group (21.31° to 63.05°).

It was noted that the knee flexion angle was dramatically reduced comparing barefoot to the orthotic condition for males and knee extension for females was dramatically increased between the barefoot to orthotic condition.

Q-angle weight bearing and supine

Q-angles in females were consistently and significantly higher than their male counterparts. This was expected as it is well documented that females have larger Q-angles (Livingstone 1999; Horton & Hall 1989).

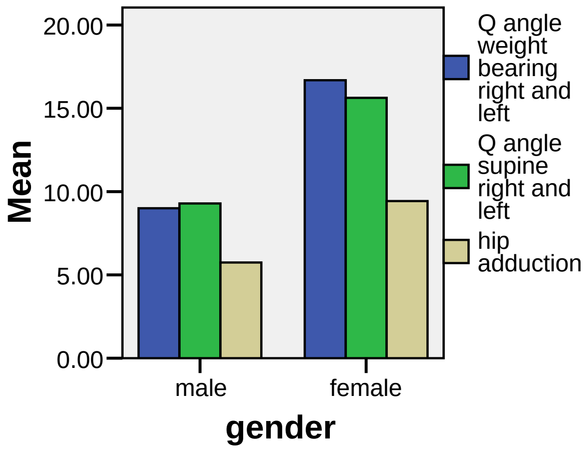


Figure 6.39 Q-angles and hip adduction in males and females

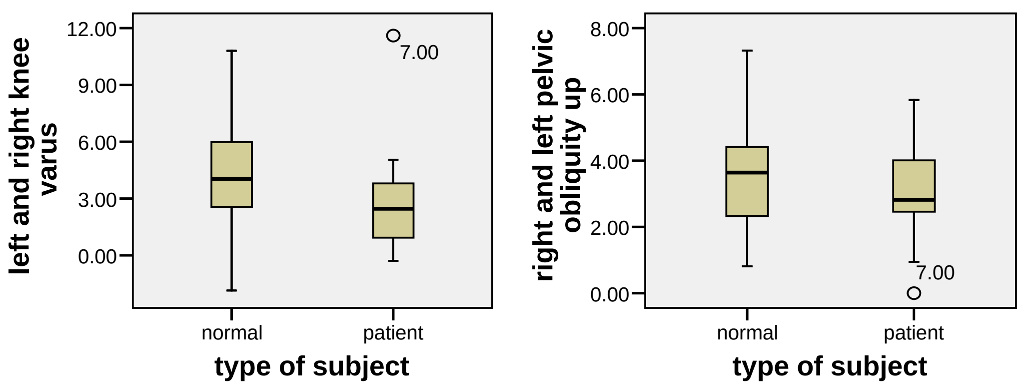
The graph clearly shows the higher values for both Q-angle positions and hip adduction in the female group. This is consistent with Hamill et al (1999) who state that greater Q-angles result in greater hip adduction values. They also state that larger Q-angles result in larger values of foot pronation. A paired sample *t*-test was conducted to test for any significance between Q-angles and rearfoot angles. Results showed there was statistical significance between Q-angle weight- bearing and supine positions and rearfoot angle (P=0.05; 0.04).

It was found in the clinical assessment that the women in both asymptomatic and symptomatic groups had higher Q-angles than the males and the symptomatic female group had higher than the female asymptomatic group. It seems that a higher Q-angle may certainly influence biomechanics of the knee joint by creating an abnormally increased valgus angle. This exerts a laterally directed force leading to mal-tracking and excessive pressure on the patellofemoral articulation, consequently resulting in anterior knee pain (Emami et al 2007).

However, an increased Q-angle cannot solely be responsible for this problem. In the control group, 33% of men had a higher Q-angle weight bearing and 13% of those had a Q-angle of over 13 degrees and 60% of females had a Q-angle of >15 degrees and one subject had an abnormally high value of 25° in the weight bearing position and 23° in the supine position. It must be recognised that despite the higher values in the normal group, nobody had any knee pain symptoms.

It was noted when looking at the waveforms of the kinematic graphs that there were a couple of unexpected patterns within the range of the graph. This occurred in many of the different variables so an extreme values test was carried out to look for outliers. In this group of symptomatic subjects there were two specific subjects who continuously had extreme values whether it was the highest or the lowest. This is shown in the box plots below.

Maximum box plots



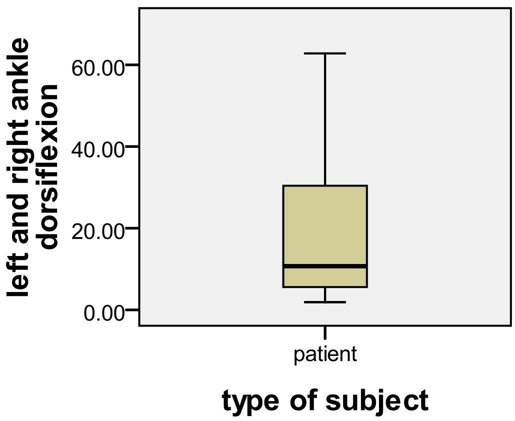


Figure 6.40 Boxplots displaying extreme values for knee varus, pelvic obliquity up and ankle dorsiflexion

The values were so extreme in some cases that it would have been a physical impossibility for a subject to replicate these measurements. The dorsiflexion graph displays a long right hand tail which suggests a positive skewness. Normally the tail is where the problem in the data lies. In this particular graph, one subject had a maximum ankle dorsiflexion value of 62.8°. This value suggests that this was nothing to do with the subjects structurally but perhaps to do with marker placement error, ghosting, and marker wobble or computer processing error. This subject's results were obviously read with caution and should not reflect any meaningful conclusions. Similarly to the outliers in chapter 5, the data was re-checked and a re-test run. In this case, the results were the same as previously which indicates that there may have been an error during marker placement or some movement occurred with the marker during the procedure. It was imperative that no data was removed from the study without due justification as it is vital to report data with and without any suspected outliers in the analysis.

Maximum kinematic values in symptomatic group

Maximum kinematic values were very similar to data from Hunt et al (2000) with the exception of one variable ankle abduction which was slightly lower but within their range of values.

In the maximum group the experimental results indicate that orthotics had a significant effect on most angles compared to barefoot. Knee extension, knee varus and external rotation of knee did not have any significant effect when comparing barefoot to orthotic condition.

No significance was found between maximum dorsiflexion but good significance was found between ankle abduction and orthotics. There were differences between barefoot to trainers and barefoot to orthotics. The results indicate that maximum ankle dorsiflexion failed to exhibit significant values however by removing outlier number 9 the results indicate that maximum ankle dorsiflexion is significant over the three conditions ($F = (1.01, 13.10) = 4.96$, $p=0.04$).

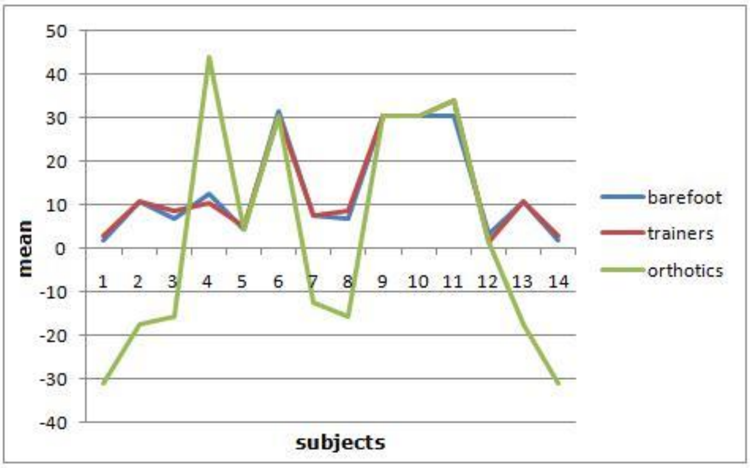


Figure 6.41 Maximum ankle dorsiflexion without the outlier over three conditions

Peak ankle dorsiflexion increased in angle from the barefoot to the trainer condition slightly but then decreased a large amount in the orthotics condition. Chen et al (2010) found that their peak dorsiflexion increased throughout the same three conditions. It was expected that ankle dorsiflexion would increase throughout the three occasions.

Branthwaite et al (2004) also found no significance between insoles and peak ankle dorsiflexion or ankle abduction. They suggested that the failure of ankle dorsiflexion being significant could be due to the fact that the angle is not a substantial component in pronation that is influenced by anti-pronatory devices.

Chen et al (2010), Mundermann et al (2003) and Stacoff et al (2000) found that maximum ankle plantarflexion angle was reduced in trials with shoes and insoles. This is consistent with the present study's results. It was found that maximum plantarflexion angle was significantly reduced when comparing barefoot to orthotics group (p=0.006) and trainer to orthotic group (p=0.009).

Previous biomechanical studies have shown that orthotic insoles improve the maximum pronation angle of the foot and internal rotation of the tibia (Eng & Pierrynowski 1994; Kitaoka et al 2002; Nawoczenski et al 1995). Maximum internal and external tibial rotation did reduce in this present study when wearing trainers and orthotics.

Knee flexion angle decreased significantly from trainer to orthotic condition and barefoot to orthotic conditions but this was not consistent with Chen et al (2010) who found increased values within the barefoot to orthotic condition. They found knee flexion angle increased significantly between barefoot to orthotic condition which the present study also found but the values were too small to be significant.

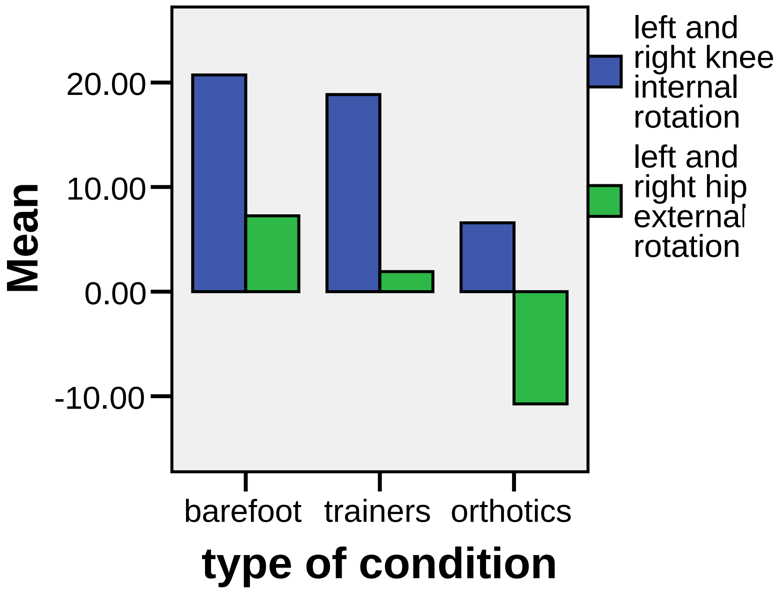


Figure 6.43 Maximum knee internal rotation without subject with extreme value over three conditions

By removing the subject with the extreme value again, the above graph clearly depicts that trainers reduce the peak knee internal rotation angle compared with barefoot and orthotics significantly reduce the angle even more than the other two conditions. Huerta et al (2009) conducted a study which examined the differences in rearfoot kinematics when using a 7 degree varus and valgus wedge however they found no significant differences between either of the wedges for peak tibial internal rotation during walking. It is however difficult to clinically correlate their results with the present study's results as they used wedges and not orthoses.

It is possible to reduce foot pronation and medial tibial rotation with a foot orthotic according to McPoil and Cornwall (2000) and Mundermann et al (2003). Recent work shows that a medially posted foot orthotic alters lower extremity kinematics during running (Mundermann et al 2003) and walking (McPoil and Cornwall, 2000). Specifically, a medially posted foot orthotic demonstrated a decrease in both the rate and degree of medial tibial rotation during walking (McPoil and Cornwall, 2000). Furthermore, given that a foot orthotic provides a mechanical barrier to pronation, the effect of a foot orthotic on lower extremity kinematic patterns should be permanent since the intervention effect is maintained.

McPoil & Cornwall (2000) state that prefabricated orthoses have been shown to be just as effective as other types of orthoses for controlling tibial rotation during functional tasks and Carcia et al (2006) found a prefabricated orthotic also reduced internal tibial rotation.

The difference between the trainers group and the orthotic group was also very significant with most of the maximum parameters except foot progression external angle and pelvic anterior tilt and pelvic obliquity up being significant. However, very few of the parameters displayed significant differences between barefoot to trainer condition. Ankle external rotation, foot progression external, knee extension and external rotation and pelvic anterior tilt and pelvic obliquity up all were significantly different between barefoot to trainers however these same parameters did not show any significance between barefoot to orthotic. Chen et al (2010) found no significant differences between the two shod conditions, trainers and trainers with insoles. Whereas it was found most of the variables were significant in these groups in this study. However, in the minimum group, there was no significance between these groups. Therefore, trainers were much the same as the orthotic group.

Comparison of maximum values between asymptomatic and symptomatic subjects

Foot progression external angle was a significant variable in maximum values. This angle measures the degree of in toeing or out toeing compared with an imaginary line on the floor. Normal values are 20° internal and 20° external values. The symptomatic group had much lower values than the asymptomatic group (5.31°- 21.35°).

It was interesting to note that barefoot knee flexion angle was higher in the symptomatic group than in the control group (mean 61.32 ° and 57.09°). An explanation for this could be that the patient group may have lacked ankle flexibility due to tight gastrocnemious (mean ankle dorsiflexion symptomatic group 10.04° and asymptomatic group 10.71°). As compensation, the subjects' may have shortened the calf muscle by increasing knee flexion and if this was still not enough they may have been forced to pronate at the rearfoot.

An increased Q-angle is often present when rotational malalignment of the femur and tibia are present. The 3D analysis system measured tibial rotation and internal tibial rotation was increased in the symptomatic group compared to the asymptomatic group. Within the symptomatic group, females exhibited the higher internal tibial rotational values over males.

It was evident that there was significant decrease of internal rotation between the groups of trainers to orthotics ($p= 0.007$) and barefoot to orthotics ($p= 0.012$). This could also match up with the coupling theory that increased knee flexion, increased internal tibial rotation and increased pronation (eversion angle) all occur synchronously. The joint angles should "in theory" then reduce with the addition of an orthotic device. It was shown that the internal tibial rotation angle decreased between barefoot to orthotics and trainers to orthotics

Internal hip rotation was also significant between the two groups. There was a slightly higher maximum mean difference between normal and injured values. It has been assumed that internal hip rotation is synchronous with foot pronation, which makes the timings of hip and rearfoot motions to be interdependent (Souza et al 2010). Values of maximum rearfoot inversion/eversion between the two groups were not significantly different although higher eversion values were noted in the patient group compared to the normal group. Similarly, values for external hip rotation were much lower in the injured group as were values for rearfoot inversion. This would indicate that might be a slight existence of a kinematic chain occurring between the hip and foot. Motion resistance exerted by the knee joint and ankle complex during the stance phase of walking may cause this chain (Souza et al 2010). Lafortune's et al (1994) findings are consistent with this and state that induced foot inversion and eversion motions during the stance phase increase external and internal rotations respectively.

Foot orthotics are typically designed to control rearfoot eversion, so they will likely reduce the relative amount of eversion to tibial internal rotation motion and thus alter their joint coupling relationship (Ferber et al 2005). However, Nawoczenski et al (1995) evaluated the effect of standard orthoses on the EV/TIR ratio of healthy runners and found an increase in the EV/TIR excursion ratio mainly due to reduced TIR. The EV/TIR excursion ratio is a single value.

6.8. Kinematic coupling between foot and leg under three conditions

The majority of biomechanical literature has reported on the kinematics of individual lower extremity joints as opposed to addressing the interaction between joints (Bates et al 1978). There are various ways to measure joint coupling relationships, such as angular excursions and joint ratios but this study looked at joint timing and joint timing differences due to its basic nature. Many have studied biomechanical coupling by examining the relationship between foot and leg using kinematic data determined at discrete points in the stance phase, for example maximum rearfoot eversion and maximum knee flexion (Nigg et al 1998 and Stacoff et al 2000).

Hamill et al (1992) investigated the effect of running in different shoes with varying midsole hardness on the timing of maximum rearfoot eversion and maximum knee flexion. The findings indicated that there were insignificant differences among the shoe conditions for time to maximum knee flexion parameter which occurred between 44.2% and 45.9% of the stance phase. In contrast, maximum eversion was found to occur significantly earlier in the stance phase (38.7%) when running with the softer midsole compared to the harder midsoles (42.8-43.8%). Subsequently, the rearfoot had started to invert while the knee was still in flexion, resulting in an antagonistic relationship with the knee. This is due to the foot imposing an external torque on the tibia while the flexing knee forces the tibia to internally rotate.

Joint timing was defined as the time to reach the maximum peak angular value and the joint timing difference was defined as a measure of synchrony between the peaks of two joint motions which normally occur during the first half of the stance phase (Bates et al 1978). It was calculated as the time to peak of the distal motion minus the time to peak of the proximal motion. The smaller the timing difference, the more synchronous relationship and a negative timing difference indicated that the distal motion reached its peak prior to the proximal motion (Dierks & Davis 2007). Measures of peak joint angles are dependent on accurate and reliable placement of markers. Slight variations in the position of the markers can result in a shift in the absolute value of inter-segment angles (Carson et al 2001), an anomaly that would alter the peak values obtained. By using the joint timing method it was understood that this method did not provide us with a description of continuous joint coupling and would therefore not reveal the complete relationship between the segments (Deleo et al 2004 and Hamill et al 1999).

The following discrete variables were indentified for each trial and subject: maximum rearfoot eversion (EV), maximum internal tibial rotation (TIR) and maximum knee flexion (KF). Eversion, internal tibial rotation and knee flexion was measured from heel strike to midstance.

6.8.1. Results

The graphs below show mean lines of all conditions and all subjects

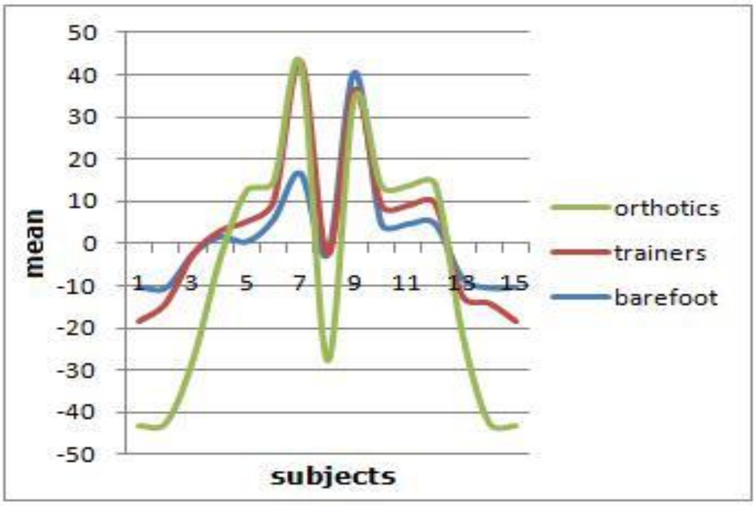


Figure 6.44 Mean values of maximum internal tibial rotation over three conditions (N=15)

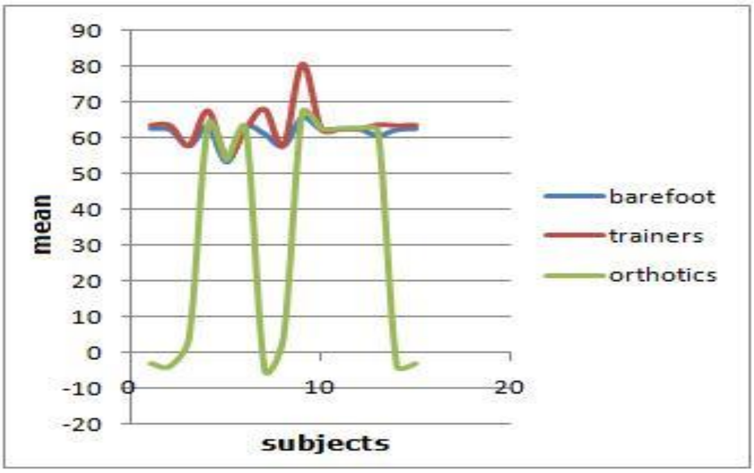


Figure 6.45 Mean values of knee flexion over three conditions (N=15)

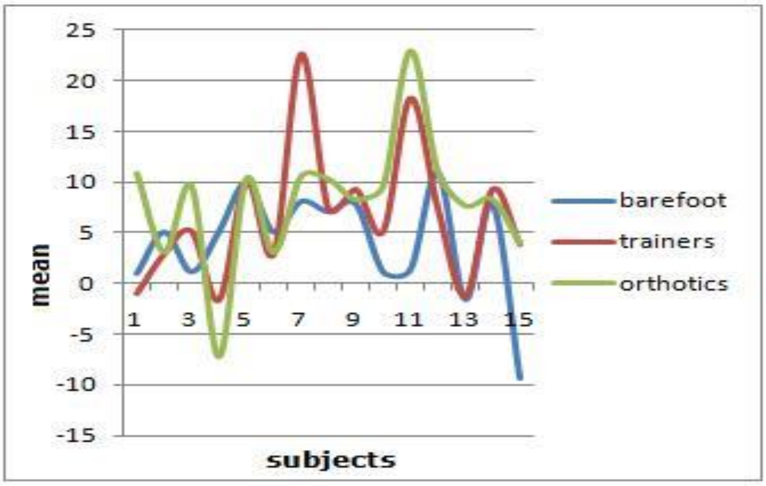


Figure 6.46 Mean values of maximum rearfoot eversion over three conditions (N=15)

The joint patterns of the maximum values of rearfoot eversion and internal tibial rotation are very similar throughout the three conditions. In both the internal tibial rotation and the knee flexion variable, all subjects except one had a lower maximum value when wearing the orthotics compared to the barefoot condition but this was not noticed when measuring the rearfoot eversion variable. However the rearfoot eversion graph showed a much more consistent pattern. It was also noticed that four subjects displayed very low values when measuring internal tibial rotation and knee flexion. This can be explained as a computer error, marker placement error or marker wobbles.

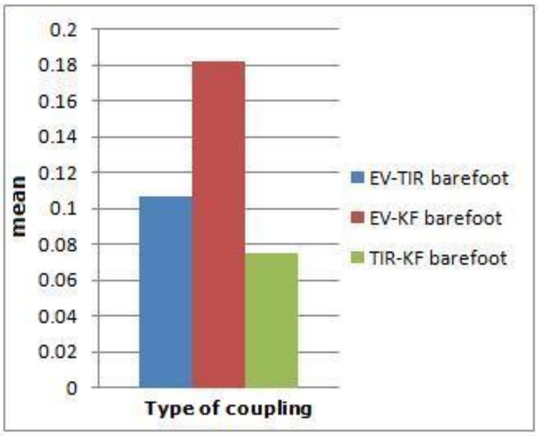


Figure 6.47 Joint timing differences for internal rotation of tibia, knee flexion and rearfoot eversion in the barefoot condition (N=15)

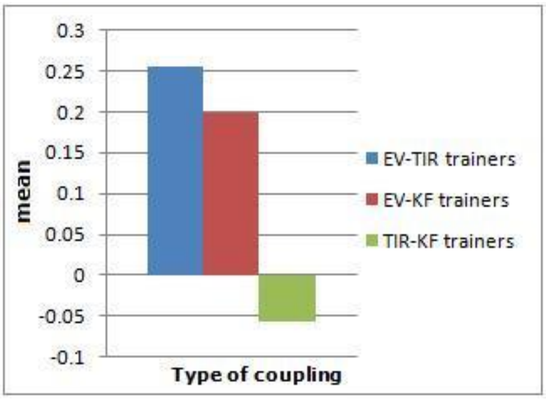


Figure 6.48 Joint timing differences for internal rotation of tibia, knee flexion and rearfoot eversion in the trainer condition (N=15)

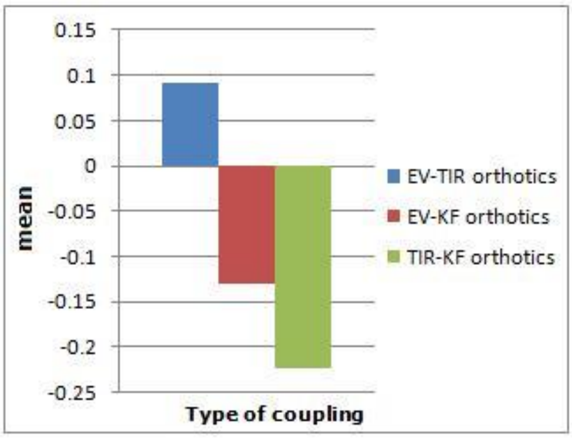


Figure 6.49 Joint timing differences for internal rotation of tibia, knee flexion and rearfoot eversion in the orthotic condition (N=15)

Based on joint timing differences, for barefoot, TIR-KF was the most synchronous compared to the other two. In the trainer group, EV-KF had the smallest value and was therefore the most synchronous. However, TIR-KF resulted in negative timing values, indicating that peak TIR reached its peak before KF. The orthotic group had a relatively synchronous relationship within EV-TIR but the other two relationships were negative. This resulted in EV and TIR peak values occurring before KF values. Dierks & Davis (2007) have also reported relatively small timing differences between EV-TIR and TIR-KF in barefoot although they also found small timing differences between EV-KF.

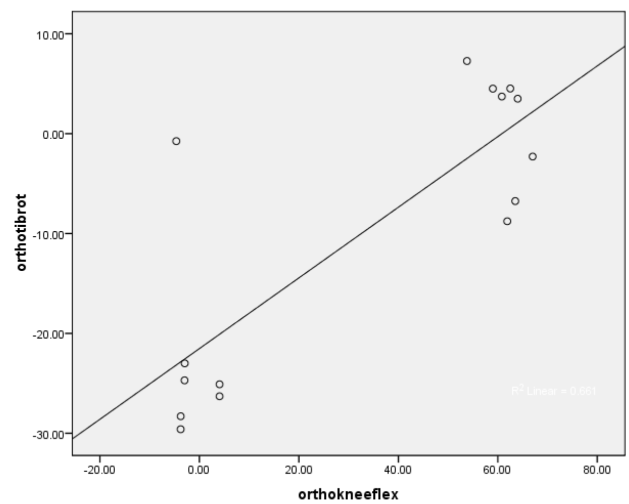
Comparisons were made of the timing differences in the three conditions using a paired *t*-test. The findings showed an insignificant change in any of the three joint timing differences in barefoot, trainers and orthotics (p=0.901, 0.164, 0.057) respectively.

Comparisons were also made between the maximum values of EV, TIR and KF during the stance phase of walking under the three conditions. Pearson correlation coefficients were used to obtain correlations between the joint angles as this examines the relationship between the two variables which are continuous in nature (Salkind 2000). The correlation values between TIR-KF, TIR-EV and KF-EV are shown in the table below

BAREFOOT COUPLING	**(R)**	**TRAINERS COUPLING**	**(R)**	**ORTHOTICS COUPLING**	**(R)**
TIR-KF	0.311	TIR-KF	-0.014	TIR-KF	0.819
TIR-EV	0.410	TIF-EV	0.719	TIF-EV	0.215
KF-EV	-0.183	KF-EV	0.084	KF-EV	0.007

Table 6.19 Correlation values (r) for TIR-KF, TIR-EV and KF-EV (n=15)

The results show that although absolute values were not comparable, the motions of TIR-EV in the trainers coupling condition (r=0.719), p = 0.03) and TIR-KF in the orthotic coupling condition (r=0.819, p<0.01) were strongly related to each other (see figure 6.50).



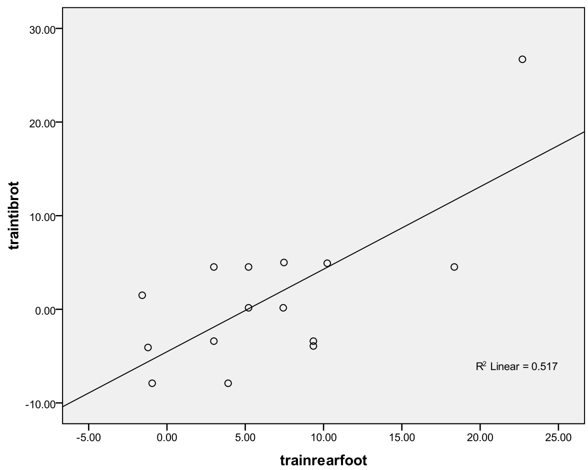


Figure 6.50 shows the positive relationship between TIR-EV in the trainer condition and TIR-KF in the orthotic condition. The regression line illustrates the line of best fit between the two tested variables in each graph (N=15).

Note the outlier in the above graph. It is mainly because of this particular value that the regression line is slightly higher. The value of the correlation coefficient is inflated because of a high influence pair of values (internal rotation 26.70° and rearfoot 22.70°).

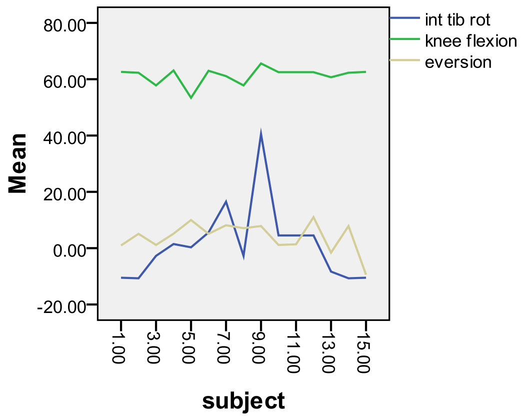


Figure 6.51 Mean degree values for internal tibial rotation, knee flexion and rearfoot angle in barefoot condition

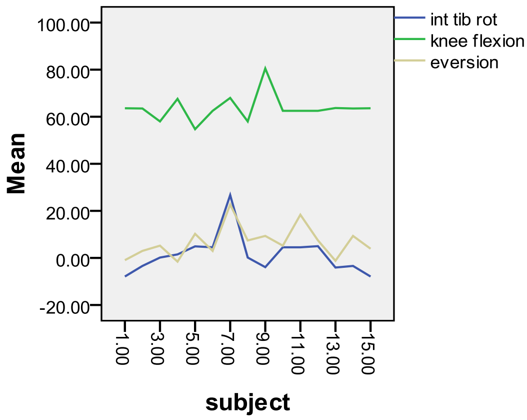


Figure 6.52 Mean degree values for internal tibial rotation, knee flexion and rearfoot angle in trainer condition

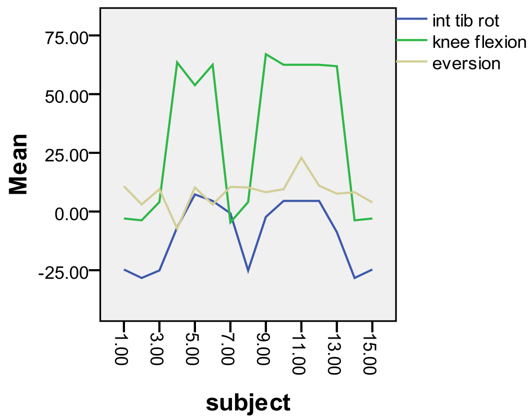


Figure 6.53 Mean degree values for internal tibial rotation, knee flexion and rearfoot angle in orthotic condition

It can be seen on the graph displaying the trainer condition that the relationship is depicting a stronger relationship between TIR and EV by looking at the joint patterns.

6.8.2. Discussion

The normal coupling of the foot and leg may be negatively affected when the timings of the motions of EV, TIR and KF become asynchronous (Dierks & Davis 2007). Tiberio (1987) suggested that prolonged EV may result in abnormal lower leg coupling and may lead to knee pain. He proposed that prolonged EV towards the end of the stance phase would be associated with prolonged TIR. During the rest of the stance phase, knee extension occurs which is associated with external tibial rotation. The tibia is internally rotating due to the prolonged EV from the first half of the stance phase so the only possible way to achieve the necessary rotation of the tibia is for the femur to internally rotate excessively relative to the tibia. This increased rotation may lead to abnormal tracking of the knee leading to patellofemoral pain.

It has been speculated that alterations in the timing between the motions of rearfoot eversion and inversion and knee flexion and extension may result in antagonistic torques being exerted at either end of the shank, thus placing excessive stress on the joints (Hamill et al., 1992; Stergiou et al., 1999; Stergiou and Bates, 1997; Stergiou et al., 2001). This is based on the assumption that both rearfoot eversion and knee flexion act to internally rotate the shank whilst rearfoot inversion and knee flexion externally rotate the shank.

The findings of this thesis suggest that although slight timing discrepancies between rearfoot eversion/inversion and shank internal/external rotation was present during walking, in general rearfoot eversion/inversion and shank internal/external rotation coupling was strong. Thus, the asynchronous timing found between the rearfoot and knee in the literature may have been due to altered kinematics between the shank and knee, rather than between the rearfoot and shank.

The findings of this research suggest that the relatively small timing differences found between EV-TIR and TIR-KF in the barefoot condition indicated that there was synchrony between these relationships. The negative values in the trainer and barefoot conditions obviously indicate little synchrony but EV-TIR values in both of these conditions indicate simultaneous peak values. It was interesting to note that there was better coupling in the trainer condition between EV-TIR compared to barefoot and orthotics. Stacoff et al (2000) found no substantial difference between calcaneal and tibial movement patterns in barefoot and shod running. Another explanation could be that the orthotics had an immediate effect on the rearfoot kinematics but did not have any effect on the joints proximal to it and was therefore "out of phase".

McClay and Manal (1998) reported that peak eversion and tibial internal rotation occurred at similar times of (0.92s) and (0.10s) after heel strike. Our findings are in agreement with this with maximum eversion occurring (0.38s) after heel strike and internal tibial rotation occurring (0.17s) after heel strike.

Bates et al (1979) reported insignificant timing differences between peak knee flexion with peak calcaneal eversion when comparing a normal group and over-pronating group during running. Cornwall & McPoil (1995) found similar correlation values (r=0.953) in the TIR-EV joint motions during walking.

Pohl et al (2006) reported EV was coupled with TIR during the first half of stance in running (r=0.917) which is slightly better agreement than this present study's correlations in any of the conditions.

6.8.3. Conclusion

The findings suggest that there was a synchronous relationship between rearfoot frontal plane eversion and transverse internal tibial rotation in the trainer condition and transverse internal rotation and knee flexion in the orthotic condition compared to the lack of coupling with the other variables. Further research need to be carried out using methods which measure continuous coupling of the angular motions throughout the whole of the stance phase and research should be conducted to ascertain how exactly rearfoot and shank motion is coupled with motion at the knee joint.

6.9. Foot model discussion

6.9.1. Navicular height marker

Static values

Navicular height measurements were tested for differences between genders as it was expected that males would have a higher navicular height than females. Although no significant differences were shown, the male group were higher in value than the female group. Zeybek et al (2008) reported a strong significant relationship between static navicular height and height of the subject (r=0.75, $p>0.05$). A bivariate correlation was carried out on these results and an insignificant indirect moderately strong relationship was found (r=-0.5, p=0.9).

Dynamic values

Dynamic measurements were quite disappointing in that the height of the navicular was reduced when wearing trainers compared to barefoot and over half of the subjects had reduced values when comparing barefoot to orthotics.

However, the values increased between the trainer and the orthotic group but this was expected (mean 15.87mm trainers- mean 19.28mm orthotics). The vertical height of the navicular tuberosity was selected as a measure of pronation (Del Rossi et al (2004) and since an orthotic is thought to reduce pronation, an increase in the height of the tuberosity would occur. Kirby (1992) states that in order to reduce or control the amount of pronation at the STJ, orthotics must increase the supination moment that is generated across the STJ axis of the foot. The most obvious was for this to occur is to have the forces generated by the orthotic directed in an upward direction and located medial to the STJ axis. Since orthotics are in direct contact with the plantar surface of the foot, the most likely method in which a supination moment is produced is by generating a force at the medial plantar surface of the foot in an upward direction (Kirby 1992).

Payne et al (2003) conducted a study using the same orthotic device and measured navicular height using digital callipers barefoot with and without the devices in a static position. There was a significant increase in the measurement value of 4.4mm when wearing the insoles. However, Payne et al (2003) attributed the change in measurement value to either supination of the rearfoot, elevation of the medial longitudinal arch or simply due to elevation of the foot. This could be due to the thickness of the orthotic which could raise the foot higher which makes making clinical conclusions difficult.

Vicenzino et al (2000) evaluated temporary felt orthotics on navicular height values before and after twenty minutes of jogging. The orthotics produced a 14% increase in navicular height before the exercise and reduced further after the exercise but was still superior to the control measurement. Del Rossi et al (2004) compared tape application and an evaluative orthotic with a 6° medial heel wedge on the effects of navicular height during exercises. Measurements were taken after application and after 15 minutes and 30 minutes of exercise. The wedge increased the navicular height significantly after 15 minutes but decreased over time. Both of these authors' conclusions were consistent however it must be noted that a reduction in the ability to control pronation of the foot such that the navicular height values return back to what they were before intervention, does not necessarily imply that the orthotic cannot continue to limit or control movement into the extreme range of pronation motion. This is where it is hypothesised that the greatest amount of tissue stress occurs (Ator et al 1991). Carcia et al (2006) state that given that a foot orthotic provides a mechanical barrier to pronation, the effect of a foot orthotic on lower extremity kinematic patterns should be permanent since the intervention effect is maintained.

6.9.2. Calcaneal Inversion/Eversion

Since inversion/eversion of the foot has been highlighted as the greatest component of supination/pronation (McCulloch et al 1993), a reduction in eversion could be directly linked to a reduction in pronation (Blake and Ferguson 1993; Landorf and Keenan 1999; Stacoff et al 2000).

Recent findings suggest a varied response in rearfoot kinematics to orthotics devices (Nigg et al 1998; Nigg et al 2003). When wearing the orthotic, it was expected that the foot would be instinctively placed in a more inverted position as Mundermann et al (2003) stated that by adding a medial post to an orthotic, the foot would intuitively be placed in an inverted position. Both barefoot and trainer conditions started heel strike in an everted position but under the orthotic condition, the results clearly display an inverted position. The results displayed from Mundermann and colleagues show a significant reduction in maximum rearfoot eversion during the first half of the stance phase when wearing a post on the orthotic. In the present study, the results were consistent with this and although the results were insignificant, there was an overall slight decrease in calcaneal eversion when wearing orthotics during the stance phase.

Branthwaite et al (2004) compared two types of insoles and found significant results of reduced eversion when comparing insoles to walking barefoot. Biplanar insoles reduced maximum foot eversion by an average of 3.1 degrees when compared to no insole condition when walking.

Stacoff et al (2000) found similar maximum inversion angles on subjects with orthotics (9.08°, SD 3.82) and subjects with trainers (8.69°, SD 3.38). In the present study the peak values for orthotics were (mean 8.09°, SD 6.32) and for trainers (mean 6.76°, SD 6.80). It was interesting to note that both studies orthotic condition, peak values decreased from the trainer condition. This demonstrates the expected effect with eversion values decreasing as orthotic correction occurs (note that positive values denotes inversion and negative values denote eversion values).

It was also interesting to note that 80% of subjects actually decreased their maximum eversion values when using the orthotics and most of the subjects increased minimum values when using the orthotics. This is not consistent with Nawoczenski et al (1995) who found that individuals with flat feet had increased eversion maximum values when wearing insoles.

6.9.3. Static vs. Dynamic

Hunt et al (2000) failed to demonstrate a significant relationship between static rearfoot angles in quiet standing and dynamic rearfoot motion. A correlation test was conducted between the static results from Vicon analysis during calibration for rearfoot motion and the results dynamically during walking. The results showed a strong relationship between them ($r=0.84$).

Cornwall & McPoil (2004) also indicate that static rearfoot motion does not influence dynamic rearfoot motion. They studied inverted and everted feet and there was no statistically significant difference ($P>0.05$) between the two groups on any of the kinematic variables measured. They seriously questioned the clinical value of rearfoot angle since it does not appear to be a good predictor of a subject’s motion pattern during walking. Further studies need to be conducted to thoroughly analyse this question however it was not the objective of this study to measure this.

6.9.4. 2D rearfoot angle

Results from the 2D angle showed that the orthotic devices did actually make a difference in reducing eversion values. Mean maximum and minimum values did show a decrease in eversion throughout the stance phase. This was very similar to the results displayed from the 3D calcaneal inversion/eversion angles.

It was interesting to note that the results from the 2D and the 3D angles were very alike. This is consistent with the study conducted by Cornwall & McPoil in 1995 who stated that 2D pattern analysis was very similar to 3D and that 2D analysis could be used effectively to assess foot inversion and eversion during the mid-stance of walking.

The rearfoot angle and navicular height and were both measured during barefoot walking using skin mounted markers, whereas they were both measured again during shod and with orthotics walking using the same markers but mounted on the shoe. When markers are attached the shoe rather than the skin, it makes it impossible to be confident that the trajectories acquired from the externally mounted markers accurately and consistently coincide with those of the internal foot structure (Chen et al 2010). It was not ideal to place the extra markers on to the last of the shoe as it was extremely difficult to accurately place the navicular marker when palpation couldn't be done to locate the top of the navicular bone.

This creates uncertainty in terms of the validity of any findings so various authors have conducted alternative methods such as Williams et al (2003) and Williams et al (2001) who put extra markers on the rearfoot by placing them directly on the heel and extended through windows cut out in the shoes. The windows allowed for unabated motion of the markers on the heel since it has previously been demonstrated that the calcaneus may move within the heel of a running shoe (Stacoff et al 1992; Van Gheluwe et al 1995). Stacoff et al (1992) attempted to quantify the effect of calcaneal slippage within the heel of a shoe during running by cutting windows in the heel counter of a running shoe so that markers could still be placed on the calcaneus. They found that rearfoot eversion excursions (angle from heel strike to maximum angle) measured using markers on the calcaneus were 1.6° different with barefoot having the higher value of eversion.

Stacoff et al (2000) used intracortical pins inserted into the foot under local anaesthetic and bone pins were then drilled into the heel and the reflective markers screwed onto the pin. Attaching the small markers to the shoe directly was the most convenient method given that the subjects all wore their own shoes. The results should be noted with this in mind.

6.10. Limitations to study

There were several limitations to this study. The foot orthotics used in this investigation were limited in that they were not individual specific in the degree of medial or lateral posting. Mundermann et al (2003) state that posting of moulded orthotics has been claimed to increase the effects of foot orthotics, therefore the "off the shelf "orthotics may not control the foot as well as a custom device.

Another limiting factor could be that not all the subjects wore the same trainers. It could be argued that a more appropriate study design should have been adopted where all the subjects wore the same footwear. However, this design is intended to imitate "real life" in that not all individuals wear the same make and model of trainers (Nester et al 2003). Arch height and navicular height were only visually measured in this study and upon conducting literature reviews after the study was performed it seems that using callipers to measure both would be more reliable (Lapointe et al 2001).

There is a lack of research regarding foot and knee motion using orthotics so this makes interpretation and comparisons difficult. Future research is warranted to evaluate the validity of using orthotics as an intervention to reduce the risk of AKP.

Another limitation could be that subjects were measured within minutes after placing the orthotic into the trainer. Whether a greater period between introducing the orthotics and testing would have made a difference is unknown although Stacoff et al (2000) reported that an orthotic may have an immediate effect on the rearfoot but not on the joints proximal to it.

Finally a limitation that has already been brought up in an earlier discussion is the foot model that is used to determine foot kinematics. There is a highlighted need to modify and develop models that can divide the foot into smaller segments for more precise understandings of how the joints in the foot behave (Pohl 2006).

6.11. Conclusions and clinical implications

From the results presented it seems clear that there are clinical and biomechanical differences between symptomatic and asymptomatic groups. Wearing foot orthoses may be advantageous to the changes biomechanically as there was evidence of small kinematic changes when wearing the inserts. It seems that the orthotic devices tested in this study presented little and small benefits in correcting pronatory changes or movements more distal to the foot. It must however be remembered that the devices used are an "off the shelf" pair and would not necessarily reflect a custom bespoke pair prescribed from a practitioner which may explain some of the variability of the results. The results agree with many previous studies that orthotics may produce small kinematic changes within the gait cycle providing the subjects have increased rearfoot eversion, increased internal tibial rotation and an increased Q-angle already present. The conclusions fulfil one of the main objectives of this chapter which was to investigate lower limb biomechanics in a group of subjects with AKP. The other objective of this chapter was to develop and test a foot marker placement model for use with Vicon 370 Kinematic Motion Analysis System. Further research is required on the model with a larger sample size and in order to compare results with other studies, running instead of walking would be advisable.

Future studies are required to determine the static and dynamic effects of orthotics on Q-angle either using a 3D motion analysis machine or radiographically using the Positional MRI measurement tool. The 2D rearfoot angle measuring inversion and eversion was also shown to be an effective measurement that can be used in a clinical environment where a 3D motion analysis machine cannot realistically be present.

The following chapter further investigates the static effects of orthotics on lower limb biomechanics under the same three conditions as this present chapter, barefoot, trainers and trainers with foot orthotics. The measurement tool used is the Upright Positional MRI Scanner. As discussed previously, it is imperative that with any measurement tool, the researcher is deemed able to carry out the measurements necessary both accurately and reliably. Therefore, a small pilot study was conducted to establish this which will be presented first in the following chapter.

Chapter 7 - An Investigation of Lower Limb Biomechanics on Symptomatic and Asymptomatic Subjects with Anterior Knee Pain Using the Upright Positional MRI Scanner under Three Experimental Conditions

7.1. Pilot study – Intrarater reliability of MRI measurements.

7.1.1. Aim

To conduct a pilot study to evaluate the intrarater reliability of measuring angles and distances using the measuring tool Osiris. Reliability refers to a condition where a measurement process yields consistent scores (given an unchanged measured phenomenon) over repeat measurements. It was imperative that the researcher, who had no experience with using this measurement program, yielded consistent values throughout the experiment in order to produce credible results.

7.1.2. Methodology

Intrarater reliability was measured for the measurements made on the MRI images using the software program Osiris software 2007.

This was done by making independent measurements on three different occasions to ensure reliability of the tester's ability to make accurate and valid measurements. It should be noted that the tester performing the measurements had no previous experience either with MRI images or with the software Osiris. An experienced senior radiographer with 10 years of experience in putting coils on patients and positioning correctly performed all the 45 scans. This was to maintain continuity and the same technique was applied to all subjects. The machine was calibrated daily to ensure reliability.

MRI Images of five normal subjects were used using the measurement programme Osiris. All three types of scan, barefoot, shod and shod with orthotic device were used.

7.1.3. Statistical analysis

Intraclass correlation coefficients (ICC's) were calculated to assess the reliability of this measurement repetition. These were the measurements taken from the same image on the three different occasions. Intrarater reliability values were established for height of navicular tuberosity, rearfoot angle, top of navicular height, medial and lateral joint space of the knee, soft tissue volume of heel fat pad, medial longitudinal arch angle, calcaneal inclination angle and talar tilt. Statistical tests were conducted using SPSS version 15 (SPSS Inc, Chicago, IL).

To satisfy the independence assumption of statistical analysis, only measurements from the right side were analysed (Menz 2004). All data were explored for normal distribution.

7.1.4. Results

Test	Type of scan	ICC Single	ICC Average
Height of Navicular Tuberosity	Barefoot	0.98	0.99
	Trainer	0.98	0.99
	Orthotic	0.99	0.99
Top of Navicular	Barefoot	0.99	0.99
	Trainer	0.95	0.98
	Orthotic	0.98	0.99
Longitudinal Arch	Barefoot	0.97	0.99
	Trainer	0.94	0.98
	Orthotic	0.98	0.99
Calcaneal Inclination Angle	Barefoot	0.93	0.97
	Trainer	0.97	0.99
	Orthotic	0.90	0.96
Talar Tilt	Barefoot	0.99	0.99
	Trainer	0.97	0.99
	Orthotic	0.99	0.99
Rearfoot Angle	Barefoot	0.97	0.99
	Trainer	0.99	0.99
	Orthotic	0.98	0.99
Knee Joint Space Medial	Barefoot	0.98	0.99
	Trainer	0.98	0.99
	Orthotic	0.99	0.99
Knee Joint Space Lateral	Barefoot	0.95	0.98
	Trainer	0.91	0.97
	Orthotic	0.99	0.99
Soft Tissue Volume Medial	Barefoot	0.90	0.94
	Trainer	0.96	0.92
	Orthotic	0.94	0.91
Soft Tissue Volume Lateral	Barefoot	0.89	0.91
	Trainer	0.90	0.87
	Orthotic	0.87	0.99

Table 7.1 Intratester Reliability of 10 variables over three experimental conditions

With respect to the 150 MRI images measured, the intratester ICC values were all very accurate with measures ranging from (ICC= 0.87-0.99). As mentioned earlier in a previous study, the acceptable ICC values are as follows, values > 0.75 indicated excellent reliability, 0.4-0.74 indicated adequate reliability and < 0.40 indicated poor reliability (Salter et al 2005).

7.1.5. Discussion

The purpose of this pilot study was to evaluate the intratester reliability of measuring angles and distances from radiographs using the measuring tool Osiris. The first consideration when interpreting the results is the reliability of the "gold standard" radiographic measurements. The test-retest reliability of all of the variables was excellent with all of the ICC values greater than 0.8. These measurement repetitions ICC's suggest a very high reliability of the method used and the consistency of the tester. These values were found to be in agreement with values reported previously (Chapter 2 section 2.8).

For comparison purposes, there are many studies which have looked at the reliability of measuring angles and distances on plain X-rays and standard supine MRI machines but very few exist which have been conducted on the Upright Positional MRI Scanner. To date there are also none which have investigated the effects of orthoses on the foot and leg using radiological scans.

However, if we compare results from previous studies using plain X-rays, it seems they are comparable. The methods may be different but the high ICC's appear to be the same when measuring the equivalent angle or distance. Most of the literature written consists of measurements of the calcaneal inclination angle (CIA) and navicular height and report excellent ICC's which is consistent with the results above.

7.1.6. Conclusion of pilot study

It was concluded that based on the measurements above, reliable measurements can be made on radiographs using the measurement tool, Osiris.

7.2. Introduction

This main study aims to look at the effects orthotic devices and footwear may have on the ankle, foot and knee joint. It will illustrate how the Upright Positional MRI Scanner can demonstrate biomechanical changes in the foot, knee and ankle between three different conditions, barefoot, shod and shod with orthoses in normal and asymptomatic patients.

In order to fully comprehend lower limb joint kinematics, a thorough understanding of it is necessary however previous kinematic studies of the leg and foot have been limited due to the invasive nature of anatomic studies but nowadays modern diagnostic and imaging procedures such as CT and MRI can make diagnosis more accurate and precise.

7.3. What is MRI?

MRI has an important role in the diagnosis and treatment of musculoskeletal injuries. It can accurately depict soft tissue injuries such as muscle, ligament and meniscal tears as well as cartilage and bone injuries non-invasively and has proven to be an excellent resource for static and dynamic joint imaging (Brown & Bradley 1994). It is a method of looking inside the body without using surgery or X-rays and provides detailed images of the body in any plane. It uses magnetism and radio waves to produce clear pictures of the anatomy of the body.

MRI is an extremely effective modality for evaluating the musculoskeletal system and because the musculoskeletal system is comparatively easy to immobilise, motion artefacts are rarely a problem. The non-invasiveness, lack of ionizing radiation and multiplanar imaging capabilities are all desirable features of MRI (Awh & Runge 2002). It uses a powerful magnetic field to align the nuclear magnetisation of hydrogen atoms in water and fat in the body. Radiofrequency fields are used to systematically alter the alignment of this magnetisation, causing the hydrogen nuclei to produce a rotating magnetic field detectable by the scanner. This signal can be manipulated by additional magnetic fields to build up enough information to construct an image of the body.

7.3.1. What is an Upright Positional MRI Scanner?

Figure 7.1 Subject in Upright Positional MRI Scanner

Traditional MRI systems are designed so that the patient is scanned in the recumbent non-weight bearing position in a tunnel with a high field magnet or between two large magnetised plates in a low-field. The Upright MRI Scanner allows patients to simply walk in and be scanned. It allows all parts of the body to be imaged in the weight-bearing position which can be the position of pain.

The knee is probably the easiest of all the musculoskeletal structures to image on an open MRI system. This is due to the cylindrical coils which can be placed around the knee which provide an improved signal to noise ratio. Furthermore, the knee is easily immobilised to prevent motion artefacts (Awh & Runge 2002).

The assessment of patellofemoral malalignment is best measured with the individual standing in the positional scanner. Patellofemoral malalignment refers to conditions where there is an imbalance of forces acting on the patella that produces abnormalities of alignment and tracking (Ellas & White 2004). This imbalance may result from a combination of variables in bony geometry, function of active and passive soft tissue restraints and functional demands. The result of patellofemoral malalignment and maltracking is unfavourable stresses and shearing forces that exceed the physiological threshold of tissues and may result in cartilage damage, degenerative changes, mechanical failure or patellar dislocation.

7.4. Advantages and disadvantages to the Upright Positional MRI Scanner

Elias & White (2004) discussed the advantages of upright MRI scanning of the patellofemoral joint compared to original supine MRI scanning and stated that the knee is usually measured in the supine resting position but pain only occurs in a loaded functioning joint. Measures of alignment will be different in the supine knee compared with the loaded knee in which most symptoms occur.

Another benefit is patient convenience. Patients can be scanned in a multitude of positions including sitting, standing, flexion and extension as well as the usual recumbent position as used in the conventional MRI scanners. They can walk in, sit or stand during the scan and walk out. They will also feel less claustrophobic due to the unobstructed view in front of the patients face.

The scanner is equipped with a motorised system which can move the patient into the magnet and place the anatomy of interest into the isocentre of the magnet.

However, one drawback to upright MRI is the increased potential for motion artefacts, especially when dealing with young children. An image artefact is not normally present but visible as a result of motion of the imaged part during the imaging sequence. This can result in a blurring or "ghosting "effect (Hornack 2000) and can produce artifactual appearances with pathology that may be misdiagnosed therefore use of immobilisation aids and clear instructions to the patient are vital.

7.5. Methodology

This experimental study took place at a purpose-built facility at Woodend Hospital in Aberdeen which houses the FONAR 0.6 T Indomitable Positional MRI Scanner. Full ethical approval from the local research and ethics committee was received prior to data collection.

MRI was performed using a 0.6 T “stand-up” Positional MRI Scanner (FONAR, Melville NY). This scanner was chosen because of its open design and the ability to image the foot, knee and ankle under the effects of gravity.

The images in Figures 7.2 and 7.3 were acquired using a solenoid coil (FONAR, Melville NY) which was placed around the ankle and knee whilst standing, allowing for imaging of the foot and ankle and knee during weight bearing.

Figure 7.2 Subject with ankle coil

Figure 7.3 Subject with knee coil

A T1 weighted sequence (TR 350 TE 20) was used. This sequence provides good contrast between bone, muscle, tendon, fat and skin, enabling easy recognition of all anatomy.

Conducting a careful screening procedure is crucial to ensure the safety of anyone who enters the area of the MRI system. Careful questioning and education of patients and volunteers helps to maintain a controlled environment and avoid potential health hazards (Rothschild and Rothschild 2000). When used properly, MRI is free of biohazards making it suitable for use with normal asymptomatic volunteers (Mitchell & Cohen 2004).

Upon entering the MRI department and prior to entering the controlled area, patients and volunteers were asked to complete a MRI safety questionnaire to ensure it was safe to scan them, in line with MRHA guidelines and departmental local rules. A senior radiographer was responsible for checking the questionnaire to ensure there were no contraindications to scanning. (See health questionnaire in appendix.) All participating subjects were given an information sheet explaining the study and gave written informed consent before starting.

All of the subjects (N = 20, 5 non-pathological and 15 pathological) who volunteered for this MRI study were also included in an earlier study using Vicon 370 Kinematic Motion Analysis System, which at the time was also situated at Woodend Hospital in Aberdeen. Although, 30 subjects completed the Vicon study, only 20 were suitable to also be involved in the MRI study due to health reasons, moving away from the area or simply not interested in participating.

The subjects in this study had therefore met all the inclusion/exclusion criteria before data collection took place. Subjects were asked to wear shorts and remove socks and shoes for the first scan.

The 9 variables which were derived from the data of the subjects' 3 scans are listed in Table 7.2

Term	**code**	**description**	**units**
Navicular height	**Nav ht**	**Distance between supporting surface and inferior surface of the navicular tuberosity**	**mm**
Superior navicular height	**Sup nav ht**	**Distance between the floor and the superior surface of the navicular bone**	**mm**
Medial longitudinal arch angle	**MLA**	**Angle between the tip of navicular tuberosity, posterior aspect of the calcaneus the most medial aspect of the first metatarsal head,**	**deg**
Calcaneal inclination	**CIA**	**Angle between the supporting surface and the inferior surface of the calcaneus**	**deg**
Talar tilt	**TT**	**Angle between two tangent lines, one to the inferior articular surface of the tibia and one to the most proximal talar contour**	**deg**
Rearfoot angle	**RF**	**Angle between the longitudinal axis of the posterior surface of the calcaneus with the medial aspect of the calcaneus**	**deg**
Knee joint space medial	**JSM**	**Angle between the medial femoral condyle and the medial tibial plateau**	**deg**
Knee joint space lateral	**JSL**	**Angle between the lateral femoral condyle and the lateral tibial plateau**	**deg**
Soft tissue volume- med and lat	**TV**	**Heel pad thickness measured by the distance from the medial and lateral calcaneal tuberosity to the plantar surface of the skin**	**mm**

Table 7.2 Radiographic measurements

7.5.1. Variables measured using Osiris Software

Figure 7.4 Talar tilt[20]

Figure 7.5 Talar tilt angle as measured on Osiris software

Talar tilt was measured as the angle between two tangent lines, one to the inferior articular surface of the tibia and one to the most proximal talar contour (Lohrer et al 2008).

[20] Lohrer et al 2008

The amount of degrees of talar tilt can indicate the amount of stability of the ankle joint (Gaebler et al 1997). There is however no standardised value for the degree of tilt at which the ankle becomes unstable. Some authors postulate that a talar tilt of more than 5 degrees compared to the unaffected side can indicate a serious ankle injury (Brostrom 1965; Smith & Reichl 1986 and Boruta et al 1990). Other authors maintain that a talar tilt of 15-30 degrees greater than the unaffected side indicates only moderate instability (Rubin and Witten 1960; Laurin et al 1968; Marder 1994 and Verhagen et al 1995).

Talar tilt values range from 0-23 degrees but most normal ankles have a tilt of 5 degrees or less (Wheeless 2009).

It was expected that the degree of talar tilt would change when wearing footwear and when wearing orthoses. Comparisons were also required to compare non-symptomatic talar tilt values with the patients with AKP to determine if one of the symptoms of AKP could be ankle instability.

Non-operative treatment of ankle instability can be addressed with orthoses or footwear modification (Johnson and Pedowitz 2006).

7.5.2. Calcaneal Inclination Angle (CIA)

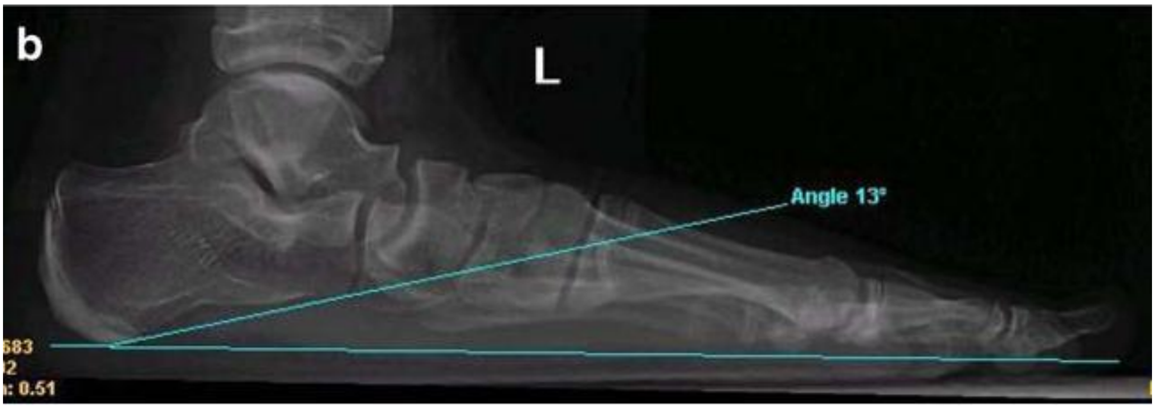


Figure 7.6 Calcaneal Inclination Angle[21]

[21] Villarroya et al (2009)

Figure 7.7   CIA angle as measured on Osiris software

The Inferior calcaneal inclination angle is a determination of arch height and is formed by the intersection of the plane of support and the calcaneal inclination axis. This axis is found from the plantar most surface of the calcaneus to the inferior border of the distal articular surface of the calcaneus (Villarroya et al 2009).

This angle is useful in evaluating a pronated or supinated foot. The more pronated the foot type, the smaller the angle, with an increased angle in a supinated foot (Donatelli 1996).

Gentili et al (1996) state normal values are between 20-30 degrees however Donatelli (1996) states the average value to be 15 degrees. Nawoczenski et al (1998) conducted a radiographic study and stated that a value below 20 degrees was a low rear foot and values above 25 degrees were considered a high rear foot. The mean values in their two groups were 16.6 degrees and 31.3 degrees.

The plane of support is determined by the most plantar aspect of the calcaneal tuberosity and the most plantar aspect of the head of the 5^{th} metatarsal (Donatelli 1996).

7.5.3. Medial longitudinal arch angle (MLA)

The medial longitudinal arch angle was first described by Norkin and Levangie (1983) as the "Feiss line"

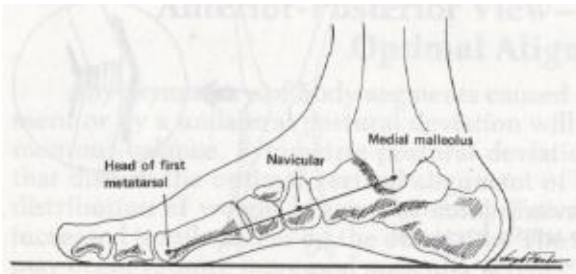


Figure 7.8 The Feiss line[22]

It is formed by a line connecting the medial malleolus to the navicular tuberosity and the most medial aspect of the first metatarsal head, and indirectly indicates the arch height (Razeghi and Batt 2002). It is suggested to compose height and length of the MLA, providing a more accurate indication of the arch behaviour (Hunt et al 1999).

[22] Norkin & Levange 1983

Figure 7.9   MLA angle as measured on Osiris software

7.5.4. Height of navicular tuberosity (arch height)

Navicular height (NH) was measured as the distance between the supporting surface and the inferior border of the navicular bone (Menz and Munteanu 2005). Palpation and measurement of the height of the navicular tuberosity has been shown to provide a useful indicator of radiographically determined navicular height (Saltzman et al 1995; Williams and McClay 2000). McCrory et al (1997) disagree and write that palpation of a bony landmark is subject to considerable inaccuracy. Palpation errors may result from an incorrect estimation of the navicular tuberosity or from other factors such as skin movement or skin distribution. They do agree however that navicular height determined by radiographic measurements is a much more exact measurement because the projection of the most inferior point on the tuberosity can be easily and reliably measured.

Figure 7.10 Navicular height measurement as measured on Osiris software

7.5.5. Top of navicular

The height of the superior surface of the navicular bone to the supporting floor was measured. This was to indicate any rotation/movement/displacement of the navicular between the three conditions. It was hoped to assess the excursion of the bone when loaded.

Figure 7.11 Top of navicular bone angle as measured on Osiris software

7.5.6. Rearfoot angle

Rearfoot angle was measured by bisecting the longitudinal axis of the posterior surface of the calcaneus with the medial aspect of the calcaneus.

Figure 7.12 Rearfoot angle as measured on Osiris software

7.5.7. Medial and lateral joint space

The knee is probably the easiest of all the musculoskeletal structures to image on an open MRI system. This is due to the cylindrical coils which can be placed around the knee which provide an improved signal to noise ratio. Coils for the shoulder are less efficient because the shoulder cannot be circled in the same manner. Furthermore, the knee is easily immobilised to prevent motion artefact (Awh & Runge 2002).

The gold standard method of measuring knee alignment is the mechanical axis of the lower limb, using weight-bearing full-limb radiographs. This technique is however, time consuming, requires special equipment, and involves significant radiation exposure (Colebatch et al 2009). It is however more commonly measured from short views of the knee which may only define limited aspects of the bones' anatomical axis (Cooke et al 2007).

The terms "varus" and "valgus" are commonly used terms throughout medical literature and confer "bowlegged "and "knock-kneed" limb deformities (Kamath et al 2010).

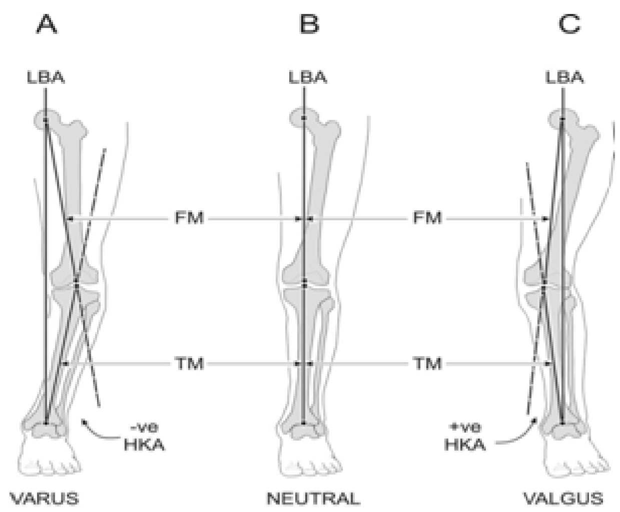


Figure 7.13 Varus, neutral and valgus positions[23]

[23] Cooke et al 2007

It was hoped to study and measure the quadriceps angle (Q- angle). However, due to the fact that this study has only short radiographs of the knee it was not possible to measure the long bones which would give us the information needed to confidently measure limb alignment so it was decided to study the medial and lateral joint space of the knee. As seen from the figure above, if the leg is in a varus position, the medial joint space will be narrowed and the lateral joint space wider and if the leg is in a valgus position, the medial joint space will be wider and the lateral joint space will be narrowed.

Medial joint space was measured as the angle from the knee centre to the medial femoral condyle and the medial tibial plateau. Lateral joint space was measured as the angle from the knee centre to the lateral femoral condyle and the lateral tibial plateau.

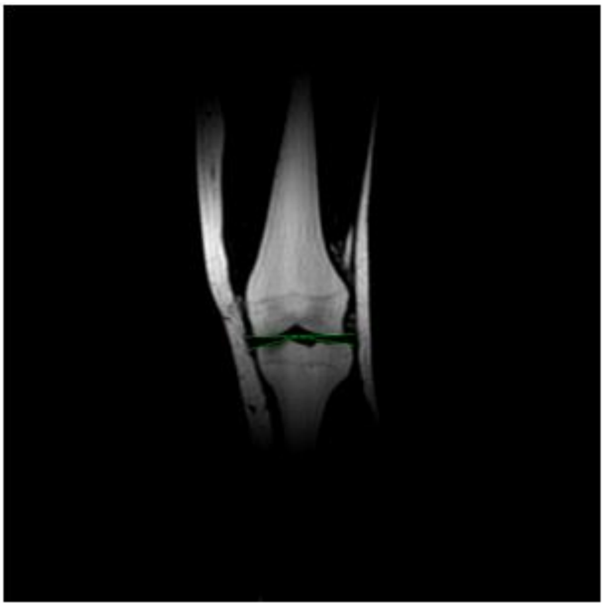


Figure 7.14 Medial and lateral joint space angles as measured on Osiris software

7.5.8. Soft tissue volume (Heel fat pad)

The heel pad consists of the fat pad surrounded by a thick sub dermal layer of fibrous tissue embedded in a layer of skin (Jahss et al 1992) with an average thickness of 18mm in the adult male (Gefen et al 2001). It is subjected to high impact forces and is responsible for the protection of the musculoskeletal system from the sudden shocks and transient forces which can propagate through the body and cause degenerative damage to the joints and also weaken the musculoskeletal system (Whittle, 2002).

There has been reported in the literature a causative link between inadequate heel pad shock absorbency and a variety of complaints such as chronic lower back pain (Volshin & Wosk 1982), Achilles Tendonitis (Jorgensen 1985) and other shock induced overuse injuries.

Trainers are designed to protect the foot yet the influence of shoes on internal foot biomechanics is not completely understood (Spears et al 2007).

Confinement of the heel due to the counter of the shoe is believed to influence heel pad biomechanics. Gefen et al (2001) measured the heel pad thickness of two 30-year-old subject's non-weight bearing as 11mm and 13mm. They then put a shoe on and the maximal deformation respectively was 3.8mm and 4.8mm. This was done dynamically and the pressures taken at initial heel strike. Rome et al (1998) measured heel pad thickness using ultrasound. They found mean measurements of non-weight bearing soft tissue volume to be 12.47mm and a weight bearing measurement to be 4.07mm.

Spears et al (2007) conducted a study aiming to measure the potential effect of confinement on internal heel pad stress during static standing and presented that confinement does have the potential to reduce stress in the heel pad in the short term.

It was hypothesised that we would see a difference in the volume of soft tissue in each of the three conditions. It was expected that the soft tissue distribution and mechanical changes with the insole and trainer would be considerably different from the unshod one.

The minimum heel fat pad thickness was measured by the distance from the medial calcaneal tuberosity to the plantar aspect of the skin and the maximum heel fat pad was measured by the distance from the most lateral aspect of the calcaneus to the skin (Spears et al 2007).

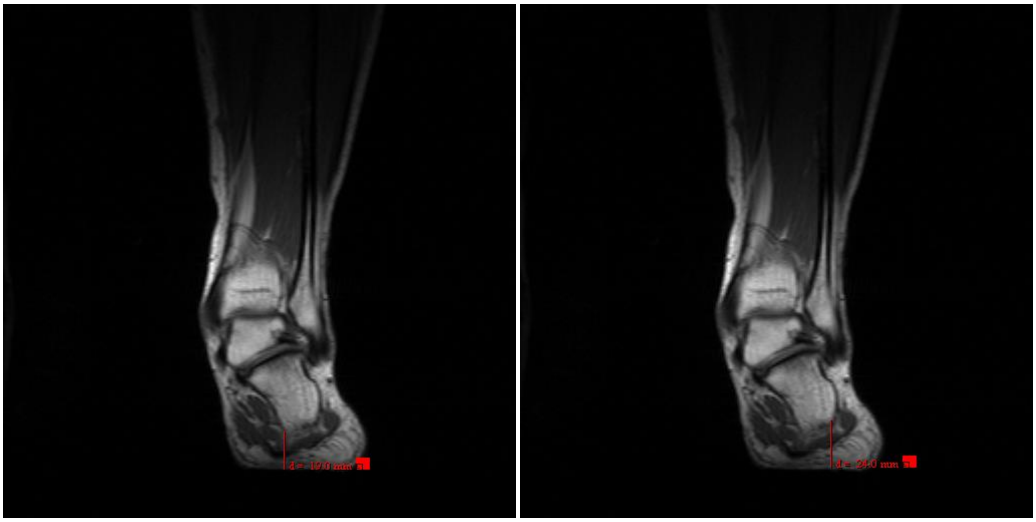


Figure 7.15 Medial and lateral soft tissue measurement angles as measured in Osiris software

7.6. Statistical analysis

All statistical comparisons were analysed using the Statistical Package for Social Sciences 15 (SPSS Inc, Chicago, IL) and consisted of descriptive tests and repeated measures ANOVAS. Descriptive statistics (mean + Std dev) were derived for either right or left legs. A repeated measure ANOVA was performed to determine if there were differences between the three conditions within the ten measurement variables. In comparing the results obtained under the three test conditions, the difference between equivalent measurements was deemed to be significant if the corresponding *P* value was (< 0.05).

When analysing differences between the ten measurements, the asymptomatic group were compared to the symptomatic group using an independent *t*-test and a post- hoc Mann-Whitney test. Level of significance was set at ($p < 0.05$). To evaluate any anthropometric related differences between the normal and asymptomatic groups, an independent sample *t*- test was used.

An extra objective of this research was to compare any associations between clinical and radiological measures. However, a full comparison of clinical, radiological and kinematic measurements is discussed in the following chapter 8.

7.7. Results

The tables below display the anthropometric differences on all individuals participating in the study

General anthropometric	Normal group	Patient group
Gender	M = 2 F = 3	M = 7 F = 8
Age mean ± SD (years)	37 ± 11.2	37 ± 9.4
Height mean ± SD (cm)	172 ± 15.8	171 ± 10.9
Weight mean ± SD (kg)	69 ± 14.8	71 ± 15.5
Left or right leg/foot	R = 3 L = 2	R = 6 L= 9

Table 7.3 Anthropometric differences between normal and patient groups

General anthropometric characteristics including age, height, weight were not significantly different between the normal and patient groups ($p>0.05$).

A full detailed display of descriptive statistics for the ten MRI measurements for normal and patient groups can be seen in the appendix. The results showed the mean and standard deviation and the range of measurement values. The largest range was the top of navicular measurement (27mm) in the control orthotics group with the minimum value of 37mm and maximum value of 64mm. The smallest range was the soft tissue volume medial aspect (4 mm) in the control barefoot group with a minimum value of 12mm and a maximum value of 16mm.

Angles and measurements are shown below for each variable in both the normal and patient groups.

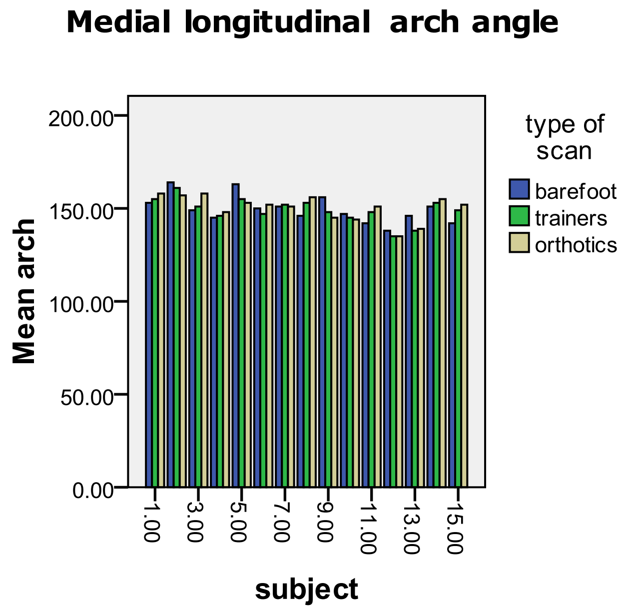


Figure 7.16 MLA patient group

(Barefoot p=0.50, trainers p=0.17 and orthoses p=0.27)

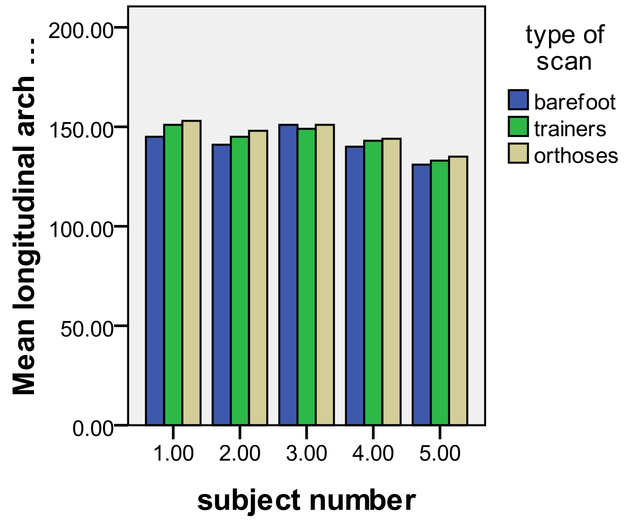


Figure 7.17 MLA normal group

In the normal group, it must be noted that most of the results showed an increase throughout the three conditions but values stayed relatively the same within the patient group which was very interesting and not what was expected. It was expected that the MLA would shorten due to the increase in height of the trainer and orthotic as most trainers have built in arch support and the orthotic device had a raised arch area which should automatically change the height of the foot

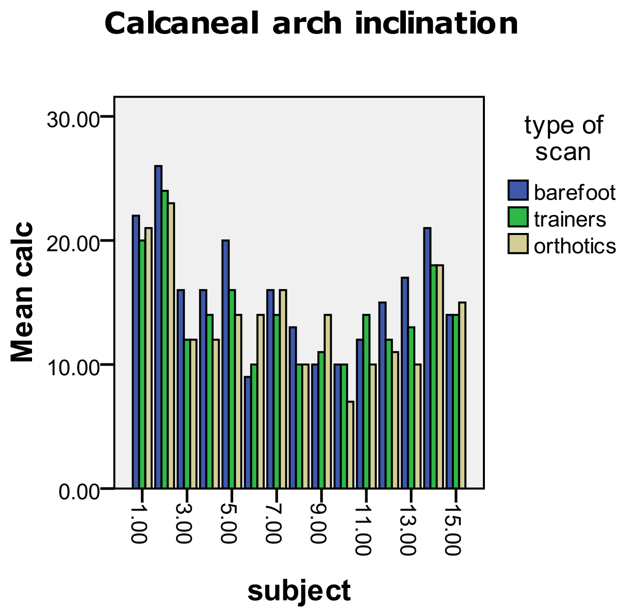


Figure 7.18 CIA patient group
(Barefoot p=0.87, trainers p=0.57 and orthoses p=0.065)

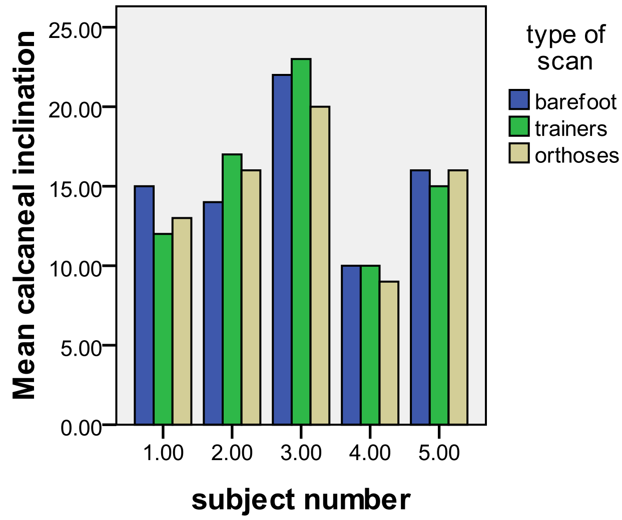


Figure 7.19 CIA normal group

In the CIA group, the angles decreased although it was only marginally - (15.4° barefoot to 14.8° wearing orthoses for normals and 15.8° barefoot to 13.8° wearing orthoses for the patient group).

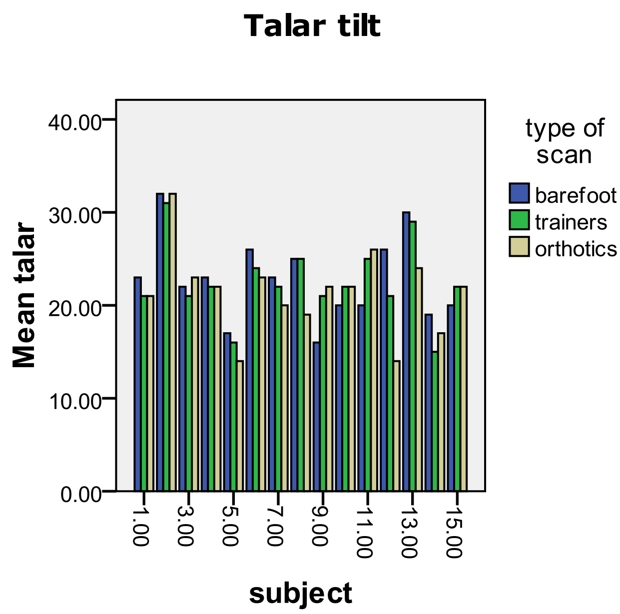


Figure 7.20 Talar tilt patient group

(Barefoot p=0.47, trainers p=0.91 and orthoses p=0.72)

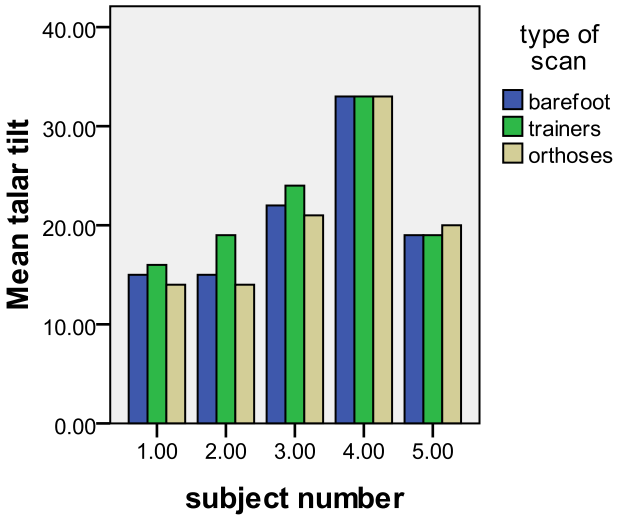


Figure 7.21 Talar tilt normal group

The results show that the patient group had some changes with the majority of patients reducing the angle throughout the three scans or simply staying the same however, the normal group did produce a lower talar tilt value when wearing orthotics but most of the group increased when wearing their trainers.

Height of navicular tuberosity

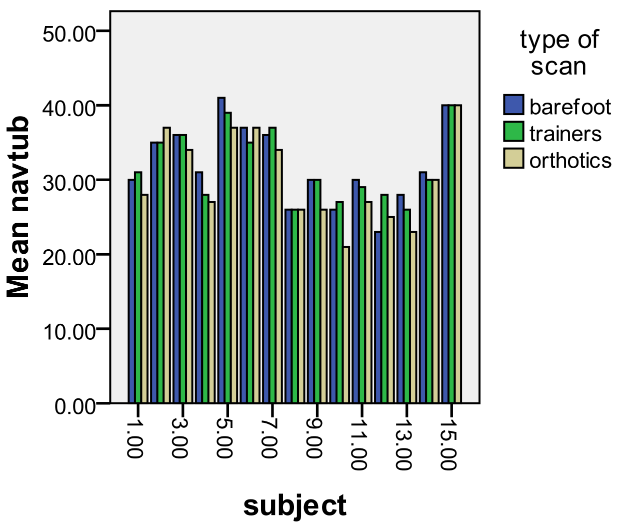


Figure 7.22 Height of navicular tuberosity patient group

(Barefoot p=0.55, trainers p=0.31 and orthoses p=0.39)

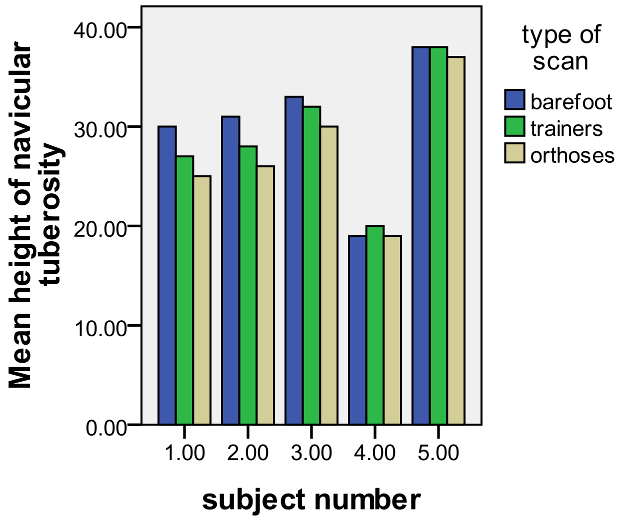


Figure 7.23 Height of navicular tuberosity normal group

Height from top of navicular

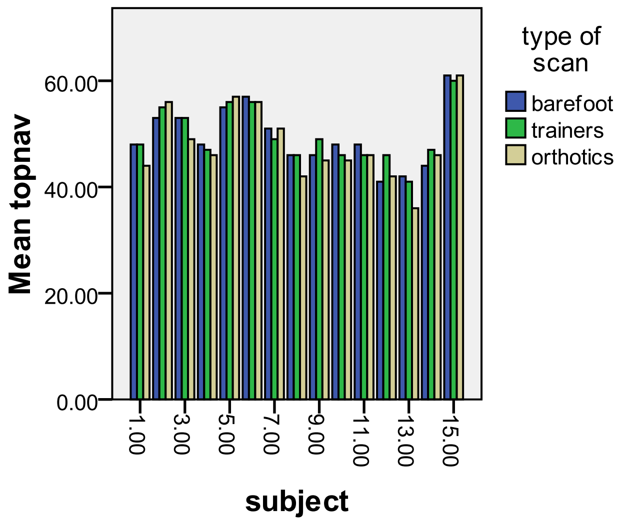


Figure 7.24 Height from top of navicular patient group

(Barefoot p=0.90, trainers p=0.93 and orthoses p=0.93)

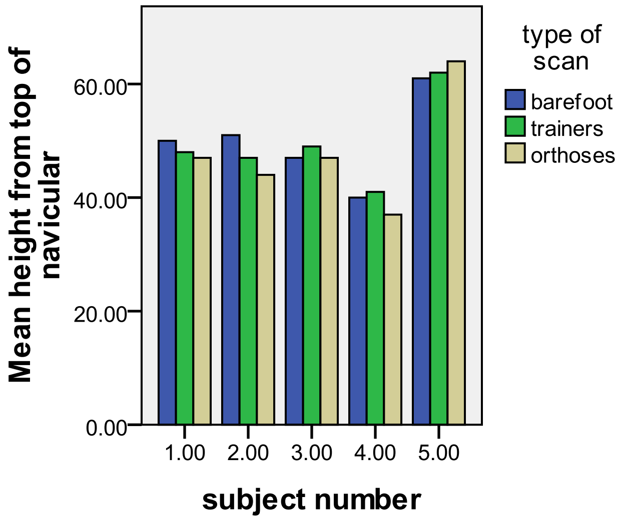


Figure 7.25 Height from top of navicular normal group

Results show that all the values except one asymptomatic subject decreased when wearing the orthoses in both groups. About half of the measurements increased when wearing trainers but decreased lower than the initial barefoot measurement when wearing orthoses and trainers. In one normal subject the navicular height decreased by 3mm from the barefoot scan to the orthotic scan.

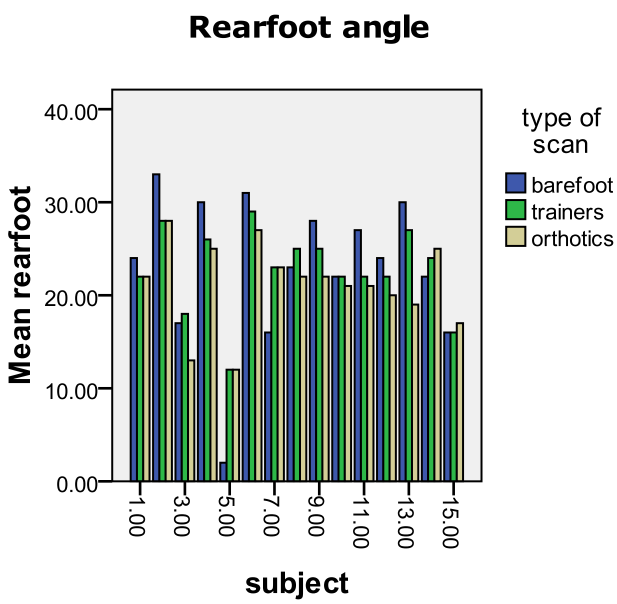


Figure 7.26 Rearfoot angle patient group
(Barefoot p=0.34, trainers p=0.15 and orthoses p=0.10)

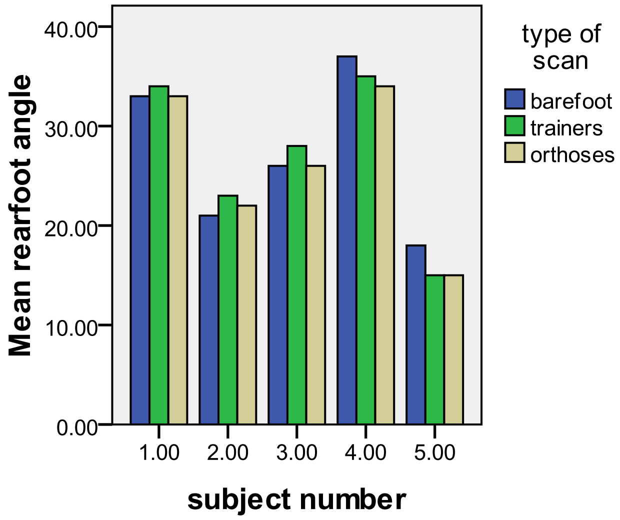


Figure 7.27 Rearfoot angle normal group

In the patient group, the results showed that most of the subjects reduced the rearfoot value when wearing the orthoses and trainers. In the normal group all of the orthotic values were less than when wearing trainers.

Knee joint medial space

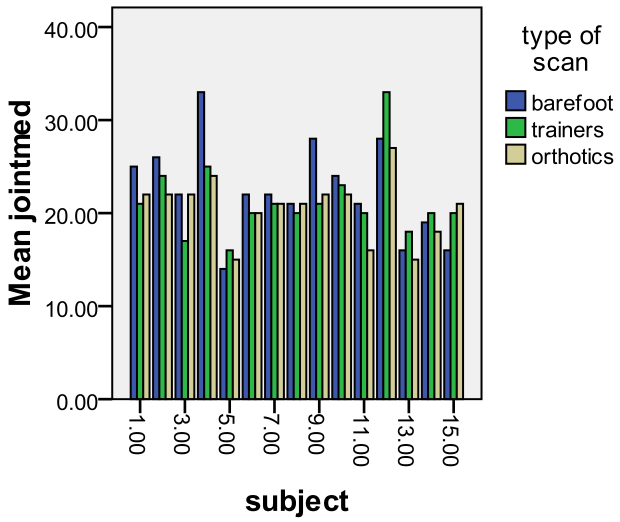


Figure 7.28 Medial knee joint space patient group
(Barefoot p=0.80, trainers p=0.65 and orthoses p=0.55)

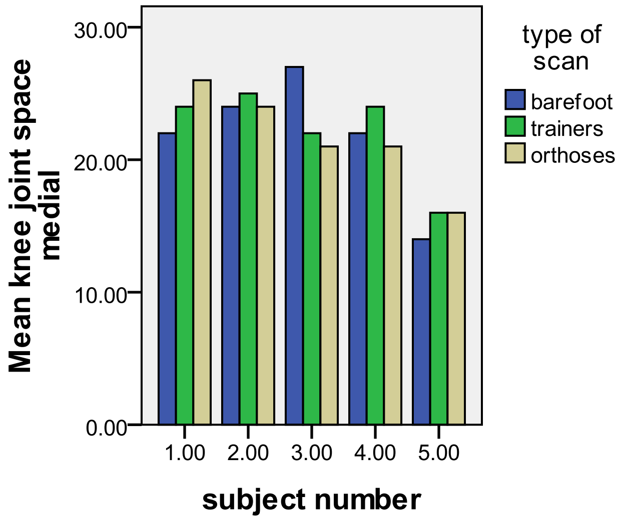


Figure 7.29 Medial knee joint space normal group

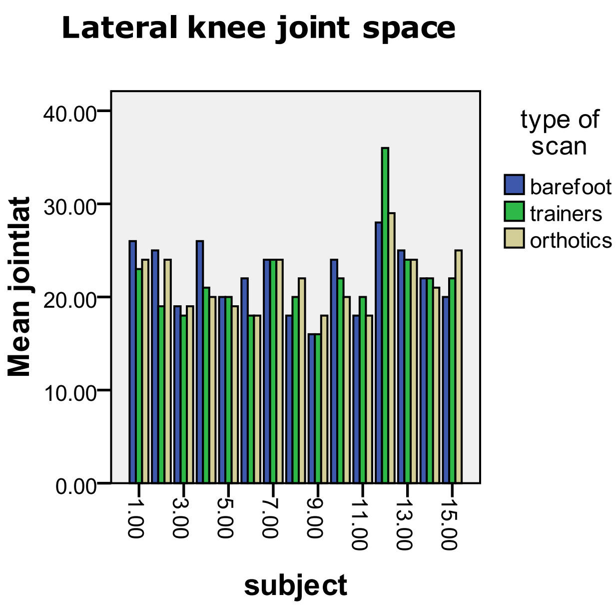


Figure 7.30 Lateral knee joint space patient group

(Barefoot p=0.92, trainers p=0.77 and orthoses p=0.93)

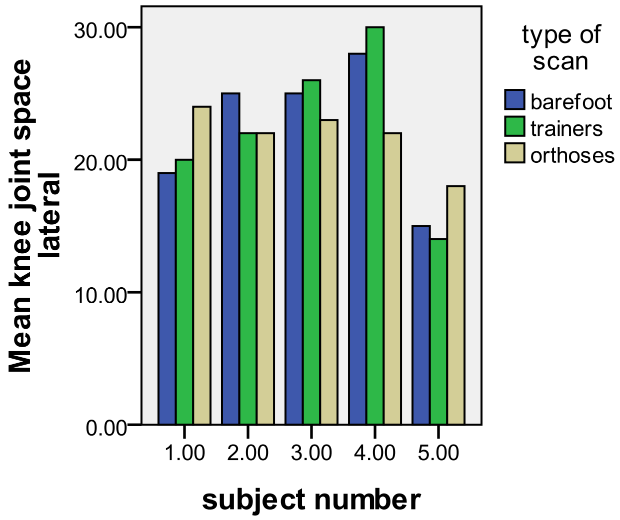


Figure 7.31 Lateral knee joint space normal group

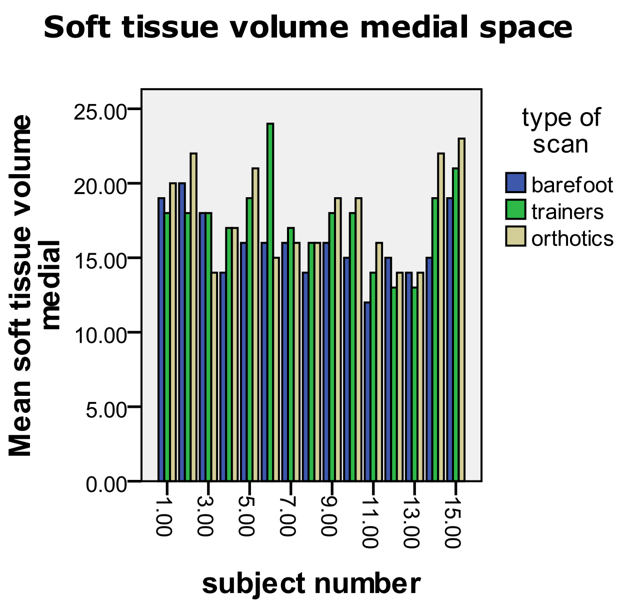


Figure 7.32 Medial soft tissue volume patient group
(Barefoot p=0.22, trainers p=0.39 and orthoses p=0.38)

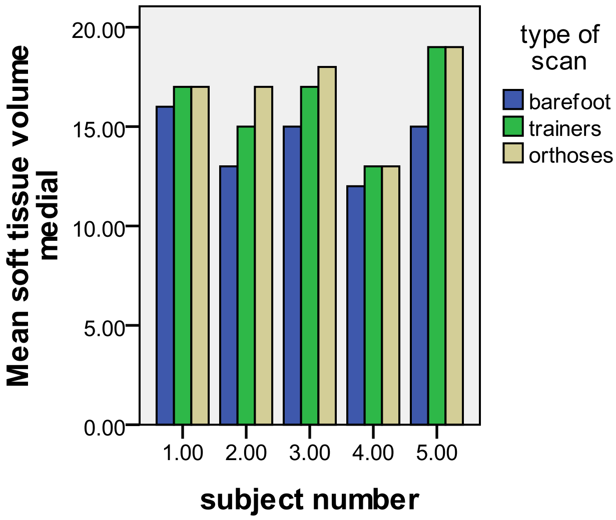


Figure 7.33 Medial soft tissue volume normal group

In the asymptomatic group, all medial values increased consecutively with each scan and the same with the lateral values with the exception of one subject whose measurement decreased by 3mm with the orthotics.

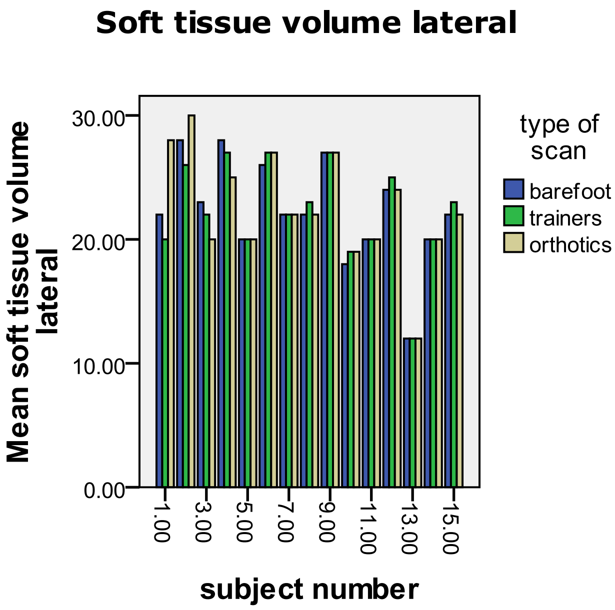


Figure 7.34 Lateral soft tissue volume patient group
(Barefoot p=0.14, trainers p=0.45 and orthoses p=0.59)

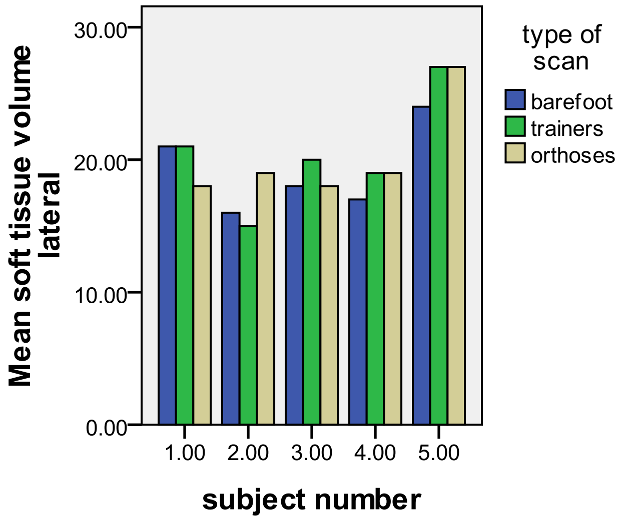


Figure 7.35 Lateral soft tissue volume normal group

A repeated measure ANOVA was performed to determine if there were differences between the three conditions within the ten measurement variables. In comparing the results obtained under the three test conditions, the difference between equivalent measurements was deemed to be significant if the corresponding *P* value was (< 0.05).

When analysing differences between the ten measurements, the asymptomatic group were compared to the symptomatic group using an independent t-test and a post- hoc Mann-Whitney test. Level of significance was set at ($p < 0.05$). To evaluate any anthropometric related differences between the normal and asymptomatic groups, an independent sample *t*- test was used.

The results of the independent *t*-test conducted to see if there were any differences between the ten variables between the normal group and patient group show that none of the p values were significant, ($p>0.05$).

P value of	barefoot	trainers	orthotics
navtub	0.55	0.31	0.39
topnav	0.90	0.93	0.93
arch	0.50	0.17	0.27
calc	0.87	0.57	.065
Talar	0.47	0.91	0.72
rearfoot	0.34	0.15	0.10
jointmed	0.80	0.65	0.55
jointlat	0.92	0.77	0.93
tvmed	0.22	0.39	0.38
tvlat	0.14	0.45	0.59

Table 7.4 Results of p values under the three conditions

When comparing the differences between the three conditions and the ten radiographic variables, the results displayed that there were no statistical significant effects ($F(18) = 0.42$ $p>0.05$).

7.8. Discussion

This study is original and to the best of our knowledge it is the first study which analyses upright MRI measurements in order to determine foot and knee alignment whilst wearing trainers and orthoses. The main purpose of this study was to determine if there were any changes in alignment of the foot and knee whilst barefoot and when wearing trainers and orthoses.

It is hypothesised that foot orthoses are used to align and support the foot and ankle complex in a more near-normal physiological position for a weight bearing foot, to prevent dysfunction or improve the function of movable body parts (Levitz

1988). It was expected to find radiological differences between the patient group and the normal group and differences between each of the three conditions.

Eslami et al (2009) conducted research in the effects of orthotics on foot angles patients with pes cavus feet during standing. They found angle changes in CIA decreased significantly by an average of 3 degrees, ($p<0.01$) when wearing orthoses (33.8° barefoot to 30.84° with orthoses) thus creating a flattening of the MLA with orthoses. This is similar to the results of this study but although the angles decreased it was only slightly - (15.4° barefoot to 14.8° wearing orthoses for normals and 15.8° barefoot to 13.8° wearing orthoses for the patient group). However, the starting angles for calcaneal inclination were much higher due to the fact that all the subjects had pes cavus (33.8°) and the majority of subjects recruited in our study had pes planus.

Murley et al (2009) conducted a study and inclusion criteria for normal arched foot for radiographic measures based on a study performed by Thomas et al (2006) to find normal values for CIA measures. They found CIA values for males to be 13.2 °- 26.2° and females 13.8° -25.6° (mean +- 1SD). The study comprised 100 adults (50 males and 50 females with a mean age of 34.3 years for males and 34.7 mean ages for females). Menz & Munteanu (2005) reported CIA mean values of 21° as did Saltzman et al (1995) who obtained the radiographic values from 100 orthopaedic patients. The mean CIA in our present study was 15.4° in the normal group and 15.8° in the patient group which is in the range of being acceptable. Both authors had larger sample groups (N=100 and 95 respectively) which may account for the different values. An important finding is that if the CIA is low, this then should correlate with the fact that the NH is lower. This may also suggest more pronated feet were being measured than supinated as the more pronated the foot, the smaller the angle; its decrease consistent with flatfoot (Villarroya et al 2009). However, it is interesting to note that CIA measurements decreased in value between barefoot and orthotics in both groups which was contrary to what was expected which was that the angle would increase due to the orthotic.

The object of an orthotic for a pes cavus foot is to improve weight distribution and stability and the orthotics all had lateral rearfoot posts which aim to move the arch in a flatter position. It would be expected therefore that the rearfoot angles measured in the study by Eslami et al (2007) would increase when wearing a lateral rearfoot posted orthotic which was reported by them. In this present study, the rearfoot angles decreased which is one of the purposes of a medially posted rearfoot orthotic (27° barefoot to 26 ° with orthotics for normals and 23° barefoot to 21.1°with orthotics in the patient group). Another differing factor is the orthoses described in their research were actually prescribed for each individual subject whereas in this present study the orthotics were off the shelf and each subject got the same pair dependent on shoe size. This was a major limitation for the study.

Vicenzino and colleagues (2000) researched whether a soft orthotics had any effect on navicular height before and after 30 minutes of exercise. They found that the subjects, all whom had at least a navicular drop of 10mm at the start of the study, produced approximately 14 % of an increase in navicular height before the exercise. They concluded that soft orthotics can help to increase navicular height which helps control abnormal pronation which should ameliorate inappropriate stresses on soft tissues of the lower limb. Payne et al (2003) found that all orthoses tested resulted in an increase in navicular height.

Del Rossi et al (2004) noted similar results as Vicenzio et al (2000). They conducted a study examining the effects of a wedged insole with a medial post underneath the heads of the metatarsals on the height of navicular and navicular drop. It was found that the insole significantly increased navicular height and navicular drop measurements.

It was also hypothesised that when subjects wore the orthotic devices, the NH measurement would increase due to the arch rising in height. The results however were surprising in that the all the values except one asymptomatic subject decreased when wearing the orthoses in both groups. About half of the measurements increased when wearing trainers but decreased lower than the initial barefoot measurement when wearing orthoses and trainers. In one normal subject the nav ht decreased by 3mm from the barefoot scan to the orthotic scan. This is unusual and not what we expected however there were other unexpected results with the MLA angles as well.

The MLA angles all increased throughout all three scans in the normal group but stayed relatively the same within the patient group. It is interesting to note that it was expected that this angle would decrease in size from the barefoot scan through to the orthotic scan (mean patient group 149.5 degrees barefoot, 149 degrees trainers, 150.2 degrees orthotics; mean normal group 141.6 degrees barefoot, 144 degrees trainers, 146 degrees orthotics). This is a confounding result and is contrary to what we expected as the purpose of an orthotic device is to reduce rearfoot motion and raise the MLA. However, it is interesting to note that although the MLA values increased when wearing the orthotic devices and the NH values decreased, the two results correlate together in that as the MLA flattens, the NH reduces which is what we would expect. The increase in arch length cannot be explained as there is no research available to back any results up with. Future research would be to re-do the study and look at the results again and compare them with the present results.

When comparing the variables, we looked at the results from the top of the navicular and compared it with the results from the height of navicular. In most cases in both groups the height and the top of navicular did simultaneous actions. When the height of the navicular tuberosity got smaller, so did the distance from the top of the navicular. In three subjects, the height of the tuberosity increased as did the distance between the top of the navicular.

Talar tilt values were expected to decrease throughout the three scans with the lower the amount of degrees, the more stability within the ankle. The results were quite surprising. The patient group had some changes with the majority of patients lowering throughout the three scans or simply staying the same however, the normal group did produce a lower talar tilt value when wearing orthotics but 100% of the group all increased when wearing their trainers. This was also seen when testing NH measurement. One of the normal subjects changed 4° between scan one and scan two. In the patient group 27% of the subjects increased with trainers but the rest all did as was expected.

There could be a couple of explanations for this. It could be the programme Osiris which was used as a tool for all the radiographic measurements is very limited in its pixel size. Clarity was often a problem when trying to locate the inferior articular surface of the tibia or other prominent areas either because of pixel limitation or through the partial volume effect around the bone margin. The subject could perhaps be moving which causes "ghosting".

Soft tissue volume was another variable which may be greatly affected by footwear. It was expected that because we were using a medially posted orthotic device, the medial (minimum) aspect of the calcaneal tuberosity to the skin would increase in distance and the lateral aspect of the calcaneus to the skin may increase slightly but should maintain the same value. In the asymptomatic group, all medial values increased consecutively with each scan and the same with the lateral values with the exception of one subject whose measurement decreased by 3mm with the orthotics. However, the medial value increased so we would presume this to be an error in measuring as the device was on a flat surface. Another explanation could be that this one subject had ill-fitting shoes or the shoes were very old and the heel counters worn down.

An important consideration which must be taken into account when we interpret the results is our sample group. We compared a relatively young group of symptomatic and asymptomatic volunteers which cannot be assumed to be equally valid for a group of older subjects. This is due to the foot undergoing several age related changes in strength (Endo et al 2002) and joint range of motion (Nigg et al 1992). Menz & Munteanu (2005) conducted a study using an older population (62-94 years) and the mean NH was 26.5mm. In our present study, the age requirement was 24-57 years and the mean for the normal barefoot was higher at 30mm and 32mm for the patient group. Other studies have also found higher values ranging from 37 to 46mm (Menz et al 2003; Williams & McClay 2000). Menz & Munteanu (2005) stated that the reason for the lower NH scores were due to 21 of their subjects had a NH measurement of less than 20mm. Staheli et al (1987) previously reported that there is an increased tendency to flat foot in people aged over 60 years. This will give the older population a lower NH measurement.

One of the main aims of this study was to assess any differences between the normal group's values and the patient group with anterior knee pain values. Variables CIA, RF, Nav ht, Sup nav ht and TV all had very similar measurements and had comparable reactions. The AKP group did have larger amounts of correction between the barefoot scan and the orthotic scan but there were no real disparity between them as hypothesised.

What was very interesting to note was the effect the trainer had on the measurements compared to the other two scans with variables of MLA and TT. One possible explanation for this could be due to cost and limited resources, it was not possible to have all volunteers wearing the same type of trainer therefore they all wore their own. These obviously varied greatly in terms of quality and shelf-life. The majority of good quality trainers have a good blend of cushioning and a medial stability post or a dual-density midsole which aids in controlling pronation throughout the gait cycle. The amount of support in a shoe could possibly affect any of these variables measured during the study and be far more supportive than the insoles under investigation. For future research, a study with the same standardised shoe would be required.

7.9. Limitations of study

Several methodological limitations are acknowledged within this study. First it must be recognised that the study was very limited to the immediate effects of the orthotic devices on any structural changes in the alignment of the foot and leg. Any permanent changes in angles or measurements may induce long term adaptation (Eslami et al 2009). Secondly, this study was conducted standing only. Dynamic measurements through gait will differ from static measurements.

One of the main purposes of this study was to measure the Q-angle with the upright MRI scanner. We had hoped to determine if there was any difference in the Q-angle between the barefoot, trainers and orthotic scans. However, as mentioned above it was not possible to measure the whole of the femur or the whole of the tibia with the MRI therefore we could not compare the clinically measured Q-angle to those measured by radiography. In 2001, Sanfridsson and colleagues did this and compared two types of radiological Q- angle measurements with a clinical measurement. The radiographic Q-angle was measured as the acute angle between the lines connecting the spina iliaca anterior superior (SIAS) and the centre of the patella and the line from the tibial tuberosity to the centre of the patella. They compared healthy knees with affected knees in semi flexion and extension. The results were surprising and they found higher values in the healthy group in both semi flexion and extension (7.3 degrees semi flexion and 9 degrees extension compared with 9 degrees semi flexion and 13 degrees extension). There was also a poor correlation between clinical and radiographic measurements ($r = 0.47$). In a similar study by Sanfridsson et al also in 2001, they found higher Q-angles in females than in males in both semi flexion and extension ($p=0.007$ and $p=0.0005$ respectively). Brattstrom (1964) reported similar findings in females and said this was due to their broader pelvis.

Relatively large pixel size and as mentioned previously, partial volume size was a limitation which has to be considered which will affect the measurement procedure. Another large problem was the difficulty in reproducing the slice position between the barefoot, trainers and orthotics conditions. It would be unlikely that the same slice was measured within each scan. Future research would be to conduct repeated scans on each condition.

Another possible idea for further research would be to use the measurements obtained from the MRI images and compare them clinically. The quadriceps angle plays a very important role in measuring alignment of the knee and it was planned to include this as a measurement however due to time restrictions with the MRI scanner it was not possible to measure the femur and hip. Therefore further research could be in measure the Q-angle using MRI and compare the results with a clinical measurement.

7.10. Conclusion

Radiographs used to assess foot and leg mechanics should be taken in a weight bearing position as to a more closely approximate relationship in gait. This will help to demonstrate how the weight bearing foot and leg reacts in a closed kinetic chain environment as they interact with ground reactive forces.

It has been shown that by doing this it is a reliable and valid measurement tool although maybe not readily available due to cost and time for many clinicians.

Chapter 8 – Comparisons of Different Clinical Measurement Methods

8.1. Introduction

3D motion gait analysis and MRI positional scanning all provide standard measurements however, measurements performed in a clinical environment by an experienced clinician are more convenient, cheaper and more accessible.

An extra objective of this study was to discover if there were any comparisons or correlations between measurements taken in a clinical environment by a clinician and the same measurement conducted by a different measuring tool such as 3D motion analysis machine or the MRI positional scanner. Measurements conducted in a clinical setting by an experienced clinician are an established and accurate way of gathering data and these results have to be compared with measurements by new methods.

In the previous chapters, each measurement tool was discussed separately and in this chapter the aim is to find out if correlations can be made from the findings from each one.

Measurement of the navicular height is commonly used as a measure of subtalar position (Holmes et al 2002). The height of the medial longitudinal arch is often used as a surrogate, albeit indirect, measure of abnormal foot pronation and navicular height which is a measure of the medial longitudinal arch of the foot, decreases with pronation of the foot (Vicenzino et al 2005). Therefore simple and reliable methods to measure static and dynamic navicular height are warranted.

8.2. Methodology

Navicular height was measured in the clinical assessment by an experienced clinician by marking the navicular tuberosity with a pen. The subject was then placed in the NCSP position and asked to maintain this position. The height of the black mark to the supporting surface was measured with a ruler and recorded in millimetres (mm). They were then asked to relax the foot and the height of the mark to the supporting surface was recorded again. Navicular drop was calculated by subtracting the two measurements from each other. This same measurement was also recorded by the 3D Motion Analysis System by placing a reflective marker on the top of the navicular tuberosity on the foot. This then enabled the height from the marker to the floor to be measured. The Positional MRI Scanner also helped measure the distance from the tuberosity to the floor and from the top of the navicular to the ground.

A repeated measures ANOVA was executed to test for a significant main effect and a post-hoc Bonferroni adjusted pair-wise comparison was used if significance was shown. To determine this degree of association between the clinical, radiological and kinematic measurements, Pearson *(r)* correlation coefficients were calculated (Menz and Munteanu 2005).

8.3. Results

Results for navicular height values when comparing three measurement techniques were that significance was shown (p=0.0001).

From the table below, there was 19.37° difference between clinical and Vicon measurements (p=0.0001), 16.54° difference between Vicon and MRI measurements (p=0.0001) and only 2.83° between clinical and MRI measurements (p=0.480).

		Mean	N	Std. Deviation	Std. Error Mean
Pair 1	clinic	34.83	15	8.10	2.09
	vicon	15.46	15	5.81	1.53
Pair 2	vicon	15.46	15	5.81	1.50
	mri	32.00	15	5.30	1.36
Pair 3	clinic	34.83	15	8.10	2.09
	mri	32.00	15	5.30	1.36

Table 8.1 Mean, standard deviation and mean standard error of navicular height measurements from clinical, MRI and 3D analysis

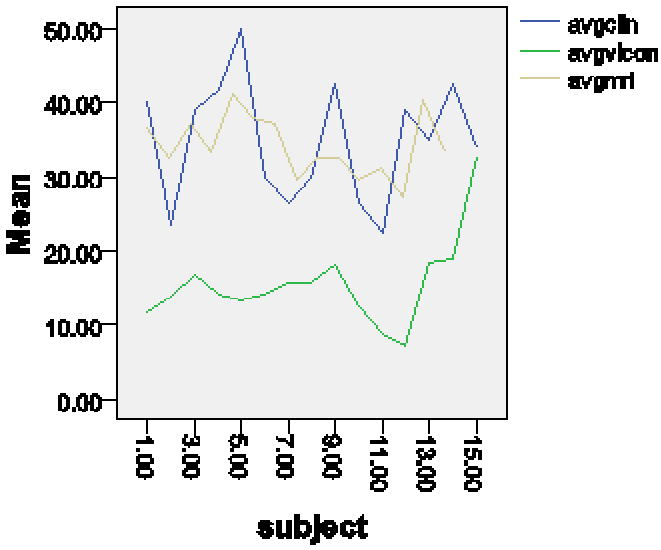


Figure 8.1 Mean values for navicular height over three different measurement techniques

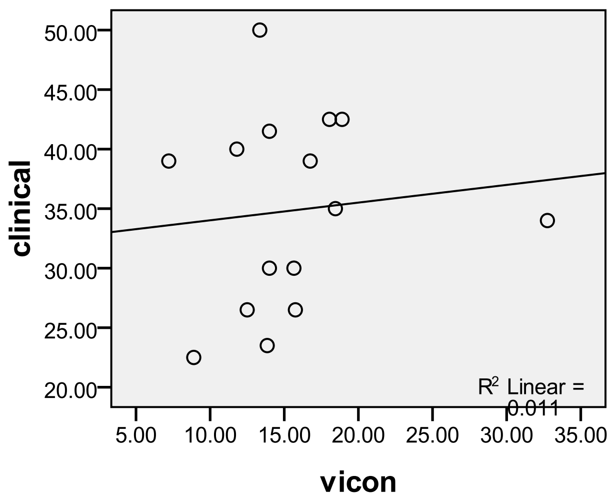


Figure 8.2 Scatterplot graph displaying clinical and 3D values for navicular height

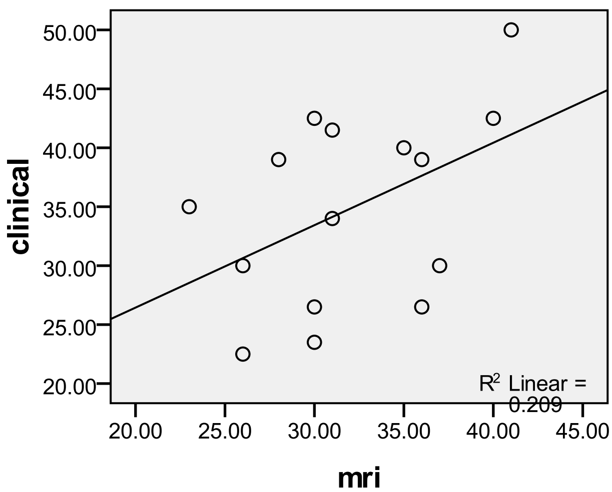


Figure 8.3 Scatterplot graph displaying clinical and MRI values for navicular height

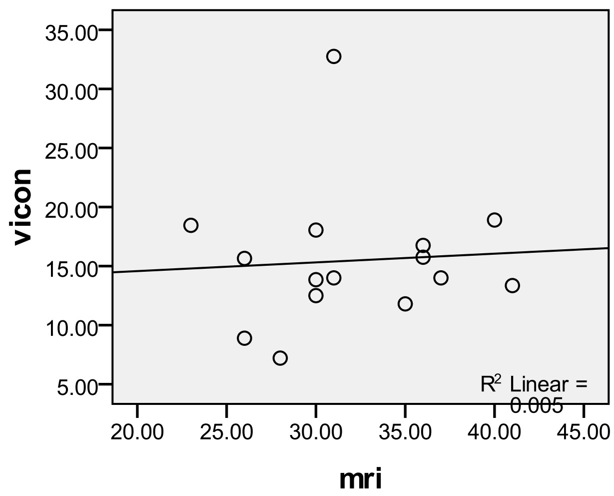


Figure 8.4 Scatterplot graph displaying 3D and MRI values for navicular height

Scatterplots showing clinical v 3D, clinical v MRI and 3D v MRI mean values for navicular height.

Scatterplot diagrams show the line Y=X which is the line of equality on which all points would lie if the three measurement tools gave exactly the same reading each time. This shows the level of agreement between measurements (Bland & Altman 1986). A matched paired *t*-test was conducted to test for correlations between the three tools and correlations were poor and insignificant.

Another variable which was measured by all three measurement tools was rearfoot angle. In the clinical assessment it was measured as the angle between a line that bisected the calcaneus and a line that bisected the lower third of the leg. When using the Motion Analysis System, two small markers were placed on the calf along the bisection line. The distal marker and the second calf marker defined two lines which were used to describe the frontal plane angle of the rearfoot angle. The MRI Positional Scanner measured the angle by bisecting the longitudinal axis of the posterior surface of the calcaneus with the medial aspect of the calcaneus.

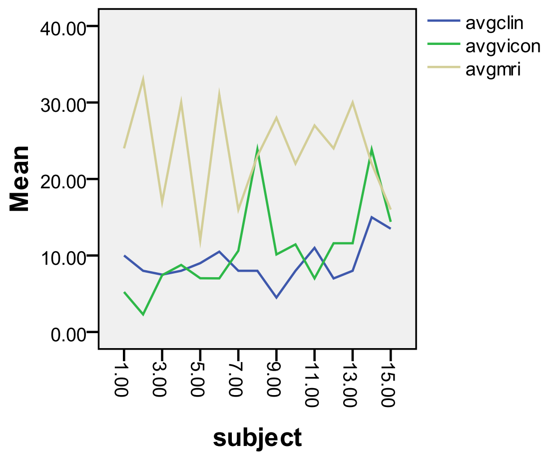


Figure 8.5 Mean values for clinical, 3D and MRI values for rearfoot angle

		Mean	N	Std. Deviation	Std. Error Mean
Pair 1	**clinical**	9.0667	15	2.61771	.67589
	vicon	10.8183	15	6.07976	1.56979
Pair 2	**vicon**	10.8183	15	6.07976	1.56979
	mri	23.6667	15	6.29815	1.62617
Pair 3	**clin**	9.0667	15	2.61771	.67589
	mri	23.6667	15	6.29815	1.62617

Table 8.2 Mean, standard deviations and mean standard of error for rearfoot angles from clinical, MRI and 3D analysis

Repeated measures ANOVA was conducted there showed a significant effect (p=0.001). The pairwise comparisons found significance between clinical measures and MRI measures (p=0.0001) and between Vicon measures and MRI measures (p=0.0001).

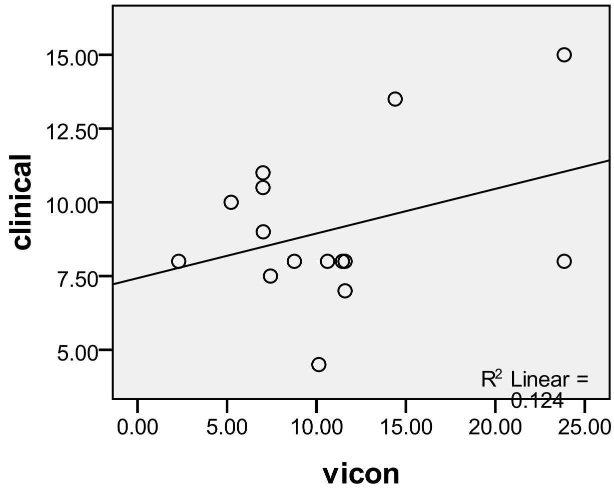


Figure 8.6 Scatterplot graph displaying clinical and 3D values for rearfoot angles

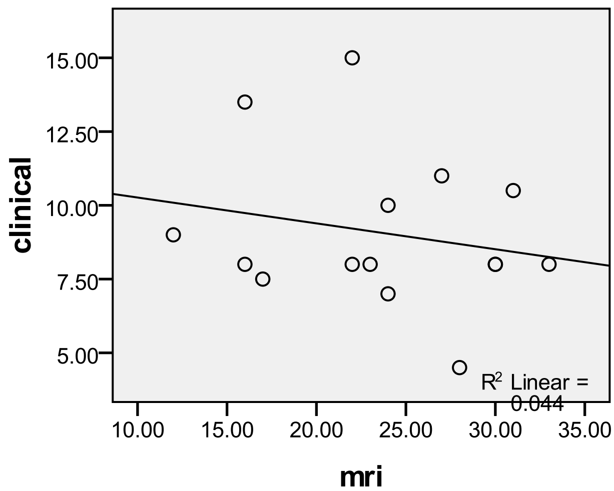


Figure 8.7 Scatterplot graph displaying clinical and MRI values for rearfoot angles

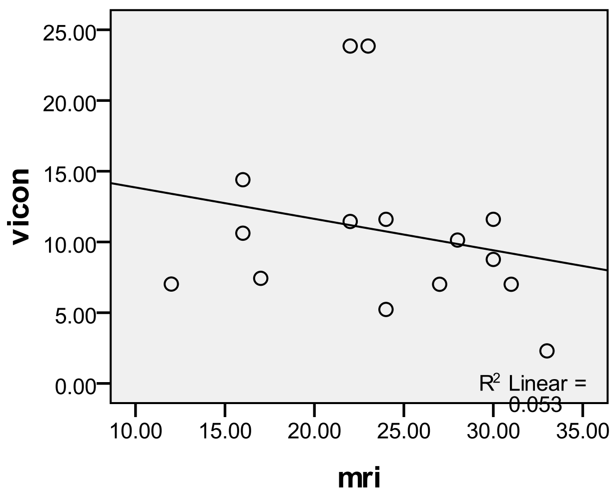


Figure 8.8 Scatterplot graph displaying 3D and MRI values for rearfoot angles

Scatterplot diagrams showing clinical v 3D, clinical v MRI and 3D v MRI mean values for rearfoot angles are shown above.

A matched paired *t*-test was conducted to test for correlations between the three tools and correlations were poor and insignificant which can be shown easily by the graphs above. They show no relationship exists between any of the three comparisons.

8.4. Discussion

Associations among measurements were compared and results demonstrated no significant associations (P>0.01) with any of the parameters. Clinical measurements of navicular height had only a small relationship with the corresponding radiographic navicular height measurement (r=0.45). This disagrees with literature which states very positive correlations between them. Menz & Munteanu (2005) writes that they found a strong correlation between clinical measurements of NH and measures from radiographs (r values ranging from 0.72-0.76). These values are consistent with Williams & McClay (2000) who reported a correlation of (r = 0.87). Associations between kinematic and clinical measurements of navicular height were not related at all. This is surprising as it was expected for radiographic measurements to be increased compared to clinical measurements. There were also no relationships between the conditions when measuring rearfoot angle. Figure 8.7 and Figure 8.8 showed negative values (r=-0.21 and r=-0.23) respectively.

The results presented that the radiographic values when measuring navicular height were indeed higher but this may be due to the fact that the measurement incorporates the skin thickness especially around the heel and foot. Each 3mm slice taken from each scan from the MRI scanner is not reproducible and to be truly comparable with kinematic data, the scans would require 3D pictures.

Kinematic values may differ between subjects and each condition due to marker wobble and placement error. These limitations have been discussed in earlier chapters. Rearfoot angle, clinical and kinematic measurements had the most agreement whereas in the navicular height measure, radiographs and clinical had the most association. The angle was slightly higher using the Vicon system than when measuring clinically. The reason for this could be that the foot model which was used for the measurements measured the rearfoot as the "whole foot". It was not able to make any distinction between the forefoot and the rearfoot. Clinically, the rearfoot angle should be measured by placing the foot into subtalar joint neutral which utilises the forefoot. For this reason the 3D analysis tool may result in a cruder value. Cadaver studies indicate that during pronation of the rearfoot, the greatest amount of motion occurs at the metatarsal-navicular level (Kitoaka et al 1997). The Oxford foot model (discussed in chapter 2 section 2.7) differentiates between the forefoot and the rearfoot and is able to provide new insight into 3D foot kinematics which is sorely required in the literature.

8.5. Conclusion

To conclude associations among measurements did not demonstrate significant associations with any of the three parameters. Clinical measurements of navicular height were only very slightly associated with the corresponding radiographic navicular height measurement. This was surprising as it was expected for radiographic measurements to be increased compared to clinical measurements. The clinical and kinematic relationship of rearfoot angle was poor but may have been better if it was not for the foot model only measuring the whole foot. Further research could be conducted using the Oxford foot model which differentiates between the rearfoot and the forefoot.

Chapter 9 – Summary of findings and conclusion

9.1. Summary of findings

The overall objective of this thesis was to gain a deeper understanding of lower limb biomechanics and to establish if there is a true relationship between anterior knee pain and lower limb biomechanics. In order to address this study, nine objectives were formulated. The following section reminds the reader of these and highlights the main outcomes of these objectives in turn.

a. To establish if a relationship exists between anterior knee pain and lower limb biomechanics.

A detailed review of the literature revealed that many researchers have attempted to establish if any relationship exists between anterior knee pain and lower limb biomechanics. This revealed that there is much debate as to whether there is or isn't. Differences between methodologies, procedures and sample sizes make comparisons very difficult. The literature revealed that excessive or prolonged STJ pronation has been associated with overuse injuries of the lower extremity during running and walking. Several researchers have also documented the synergistic relationship between STJ pronation and prolonged internal tibial rotation during walking and running. In this research a small relationship was revealed between knee pain and lower limb biomechanics. It was found that in both the normal and patient groups, women had higher Q-angle values than the males and the female patient group had higher values than the females in the normal group. This could signify that a higher Q-angle may influence biomechanics of the knee joint by creating an abnormally increased valgus angle. Another conclusion was that there was a strong relationship between eversion and inversion with internal and external tibial rotation of the leg. This coupling relationship was discussed in chapter 1 and the literature revealed that a relationship, albeit small, may exist between the two. Further research is required to measure continuous coupling of the angular motions throughout the whole of the stance phase. This will ensure exactly how the foot and leg motions are coupled with knee joint motions.

b. To analyse the intratester reliability of a detailed clinical assessment.

In order for a clinical measurement to have any scientific credibility, a reliable and valid clinical measurement technique must be employed. Reliability can be defined as the amount of agreement between successive measurements of the same joint by the same tester or different testers, namely, intratester and intertester reliability respectively. Chapter 2 highlighted a detailed review of the literature available on reliability of clinical measurements and chapter 3 presented a thorough and clear procedure of how the clinical measurements were conducted. This was to ensure that the same method was conducted on each occasion in order for the researcher to prove that the method was indeed reliable. Chapter 3 also presented a small pilot study which showed the researcher carrying out the same clinical assessment three times on three separate occasions on five random asymptomatic subjects. It was concluded that with the exception of ankle plantar flexion (ICC 0.37), the results showed good to excellent reliability (ICC >0.72). It was also documented that other authors have also showed poorer results for ankle plantar flexion in their studies. This small study fulfilled the objective that a relatively high degree of intratester reliability can be expected when using the standardised protocol as described above.

c. To develop and test a foot marker placement model for use with the Vicon 370 Kinematic Motion Analysis System to analyse the biomechanics of the lower limb.

A detailed review on the measurement tool above showed that there was a lack of research already conducted on the foot dynamically. Quite a lot of the research looks at the foot as a whole instead of looking at the foot in various sections such as rear foot and mid foot. A lot of the previous studies conducted also measure the STJ in 2D which does not represent the STJ as a tri-planar joint. A review of the literature available on past and present foot placement models for use with various 3D kinematic motion analysis systems was presented in chapter 2. It was evident that the foot model which was currently being utilized by the university at the time of this research was unable to measure STJ pronation and supination, rearfoot angle and height of navicular bone. Therefore, a new foot model was developed which could measure these specific variables. This foot model was based on the original Helen Hayes foot model but the rearfoot segment was recreated with the addition of a new marker which ensures that 3D analysis could be represented. An additional marker was placed on the top of the navicular bone which would enable the system to measure the bones movement in relation to the floor and an additional 4 markers were placed on the rear of the lower leg and heel to measure rearfoot motion. It must be noted that future research would be to develop this particular model further to separate the foot into midfoot and rearfoot segments.

In order to produce reliable and repeatable results, there is a need to minimise the problems associated with the re-application of markers on different test sessions. This is why it is important to measure the tester's performance in marker application. This ensures that the confidence with which any kinematic 3D results can be used is established. A small pilot study was conducted and the results showed that there was excellent reliability for the results within the three occasions conducted on the same day as well as the three separate occasions on three different days.

d. To develop and establish values for lower limb biomechanics in a group of asymptomatic subjects using a detailed clinical assessment and the Vicon 370 Kinematic Motion Analysis System.

This objective was very important in that it is therefore very important to have a "normally distributed" database of measurements which what to compare the "abnormally distributed" to. With the two small pilot studies conducted previously showing good to excellent reliability for clinical assessment and marker positioning this objective could be fulfilled. However, there was one subject within the group who skewed the normality results and therefore could not be used in the study. Significant differences were found between genders but this was expected as there are distinct gender differences in skeletal dimensions, the walking patterns of the two genders are likely to differ. The results presented that most of the group of thirty asymptomatic volunteers could be classed as "normal" and it is therefore a justifiable database on which to compare symptomatic subjects against.

e. To develop and establish values for lower limb biomechanics in a group of symptomatic subjects with anterior knee pain using a detailed clinical assessment and the Vicon 370 Kinematic Motion Analysis System.

This objective was conducted in a similar way to the previous one but involved symptomatic subjects. The results obtained from this objective were compared against the results from the asymptomatic group. The new foot model presented that height of navicular was higher in males which is consistent with the clinical assessment results. The findings of the coupling relationship suggest that the relatively small timing differences found between EV-TIR and TIR-KF in the barefoot condition indicated that there was strong synchrony between these relationships.

f. To analyse the effect of lower limb alignment in shod subjects with anterior knee pain using the Vicon 370 Kinematic Motion Analysis System.

g. To analyse the effect of foot orthoses on lower limb biomechanics in subjects with anterior knee pain using the Vicon 370 Kinematic Motion Analysis System.

It was found in the clinical assessment that the women in both normal and patients had higher Q-angles than the males and the patient female group had higher than the female normal group. It seems that a higher Q-angle may certainly influence biomechanics of the knee joint by creating an abnormally increased valgus angle.

When looking at kinematic values in the maximum group the experimental results indicate that orthotics had a significant effect on most angles compared to barefoot. The difference between the trainers group and the orthotic group was also very significant, however very few of the parameters displayed significant differences between barefoot to trainer condition.

The findings within the coupling relationships suggest that there was a very small synchronous relationship between rearfoot frontal plane eversion and transverse internal tibial rotation in the trainer condition compared to the lack of coupling with the other variables. The negative values in the trainer and orthotic conditions obviously indicate little synchrony but EV-TIR values in both of these conditions indicate simultaneous peak values. It was interesting to note that there was better coupling in the trainer condition between EV-TIR compared to barefoot and orthotics. Dynamic measurements were quite disappointing in that the height of the navicular was reduced when wearing trainers compared to barefoot and over half of the subjects had reduced values when comparing barefoot to orthotics. However, the values increased between the trainer and the orthotic group (mean 15.87mm trainers- mean 19.28mm orthotics). One of the most important findings was that although the results were insignificant, there was an overall slight decrease in calcaneal eversion when wearing orthotics during the stance phase. This supports the hypothesis that calcaneal eversion would be reduced when wearing the orthotic device. Further research need to be carried out using methods which measure continuous coupling of the angular motions throughout the whole of the stance phase.

h. To investigate lower limb alignment on asymptomatic subjects using barefoot, shod and orthotics conditions on the Upright Positional MRI Scanner.

i. **To investigate lower limb alignment on symptomatic subjects with anterior knee pain using barefoot, shod and orthotics conditions on the Upright Positional MRI Scanner.**

These final two objectives will be summarized together. It was expected to find radiological differences between the patient group and the normal group and differences between each of the three conditions of barefoot, trainers and orthotic devices. It was found that angle changes in CIA decreased (15.4° barefoot to 14.8° wearing orthotics for normal group and 15.8° barefoot to 13.8° wearing orthotics for the patient group.). This was contrary to what was expected which was that the angle would increase due to the orthotic.

Rearfoot angles decreased which is what was expected as it is one of the purposes of a medially posted rearfoot orthotic (27° barefoot to 26 ° with orthotics for normal group and 23° barefoot to 21.1°with orthotics in the patient group). It must be remembered that the orthotic devices used in this study were "off the shelf" and each subject got the same pair dependent on shoe size. This was a limitation for the study. Further research could be to use a bespoke pair of orthotics (individually prescribed) to compare differences between the conditions. This may give a more precise account of whether the device is accurately doing what it was prescribed for.

The results of the NH measurement and MLA angles were surprising and contrary to what were expected. It was hypothesised that when a subject wore the orthotic devices, the NH measurement would increase due to the arch rising in height. The results presented with all values except one asymptomatic subject decreasing when wearing the orthoses in both groups. About half of the measurements increased when wearing trainers but decreased lower than the initial barefoot measurement when wearing orthoses and trainers. The MLA angles were expected to decrease in size when wearing the trainer and the orthotic but they all increased throughout all three scans in the normal group but stayed relatively the same within the patient group (mean patient group 149.5 degrees barefoot, 149 degrees trainers, 150.2 degrees orthotics; mean normal group 141.6 degrees barefoot, 144 degrees trainers, 146 degrees orthotics). This is a confounding result and is contrary to what we expected as the purpose of an orthotic device is to reduce rearfoot motion and raise the MLA. Future research would be to re-do the study and look at the results again and compare them with the present results. To conclude the summary, Variables CIA, RF, Nav ht, Sup nav ht and TV all had very similar measurements and had comparable reactions. The AKP group did have larger amounts of correction between the barefoot scan and the orthotic scan but there were no real disparity between them as hypothesised. Further research is required with a larger subject sample and bespoke insoles. Controlling the type of trainer would also aid in comparing the results and comparing walking with running would help clinicians understand more about the dynamic foot.

9.2. Conclusion

The purpose of this research was to investigate if a relationship exists between anterior knee pain and lower limb biomechanics. The results from the study reveal that a small relationship does exist between them. Two small pilot studies were conducted to show scientific credibility, reliability and repeatability. The first study examined the tester's ability to accurately perform the measurements in the clinical examination and revealed a high degree of intratester reliability when using the standardised protocol as discussed. The second study determined the test-re-test and intratester reliability of marker placement using the 3D gait analysis system. Results revealed acceptable results which were comparable with previous investigations for intratester repeatability of temporal-spatial and kinematic parameters. This set the precedent for the reliability of any clinical measurements conducted throughout the study.

A foot model was devised based on the Oxford foot model as there is a definite need for a standardised multisegmental foot model. Most previous foot models involve the rearfoot only and as the foot is extremely complex, more information is required to gain more knowledge of how the foot works both statically and dynamically. For the purpose of this research, rearfoot inversion and eversion and navicular height were able to be measured with the foot model. Further work is required to create a model which differentiates between the rearfoot and the midfoot.

Two groups of subjects were used, an asymptomatic group and a symptomatic group. A study was conducted on the asymptomatic group to determine a normative database for which any other results can be compared against. The normative database was then used to compare the results from the symptomatic group. The group with anterior knee pain were measured under three experimental conditions, barefoot, trainers and trainers with orthotic insoles. Any biomechanical changes were noted when analysing the 3D kinematics and temporal parameters during the three conditions. The results show clearly that there are clinical and biomechanical differences between the two groups. There was small evidence which showed that wearing foot orthotics may be advantageous to the changes biomechanically. There were little and small benefits in correcting pronatory changes or movements more distal to the foot when wearing the orthoses. This compared well with previous authors who state that orthotics may produce small kinematic changes within the gait cycle providing the subject has increased rearfoot eversion, increased internal tibial rotation and an increased Q-angle. A large limitation of this study was that Q-angle was not measured using 3D analysis. This is a future recommendation as would provide a more detailed and valuable insight into dynamic foot function during walking.

The final section of the study was to investigate the static effects of barefoot, trainers and orthotics on lower limb biomechanics using the upright positional MRI scanner. A small pilot study was carried out to evaluate the intratester reliability of measuring angles and distances using the measurement tool, Osiris. It was concluded that the results of the intratester reliability were excellent. However the results of the MRI study were slightly disappointing. There were many limitations throughout this section of the research such as pixel size and the coil size used.

A small comparison study was carried out to compare 3D kinematic analysis, clinical examination and MRI positional scanning. It was expected that there would not be any correlations between the three measurement procedures.

However, significant associations were found between clinical navicular height and the corresponding radiographic navicular height measurement. Associations between kinematic and clinical measurements were not as strongly related as the relationship between radiographic and clinical measurements. However further research is required to be done with a new multisegmental foot model which may change the associations with clinical measurements.

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Appendices

1. Poster for subject recruitment for reliability study
2. Poster for subject recruitment for 3D kinematic analysis
3. Asymptomatic subject information sheet for 3D kinematic analysis
4. Symptomatic subject information sheet for 3D kinematic analysis
5. Clinical Assessment form
6. Consent form
7. Foot model marker (mkr) and model (mod) files for foot model
8. MRI health questionnaire sheet for asymptomatic subjects
9. MRI health questionnaire sheet for symptomatic subjects

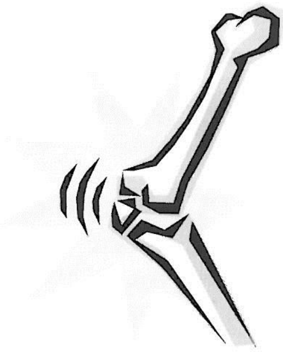
Got 2 hours to spare to help a fellow student?

Are you are a 1st year Physiotherapy student aged 17 to 25 and interested in participating in an intratester reliability study assessing clinical measurement of the lower limb?
All students should be free of pain, musculoskeletal and neurological dysfunction of both right and left lower extremity, and no intervention of foot orthoses. In addition, none of the students should have been treated for musculoskeletal disorders of the right or left lower extremity in the past year.
If interested please write your name below, contact myself at PH 113, leave a note in my pigeonhole or alternatively contact Elizabeth Hancock.

Thank you for your help

Gillian Duncan
PhD research student

GOT KNEE PAIN?



AGED 18-40?

Interested?

For more information please contact:
Gillian Aitkenhead
Research Student
Robert Gordon University
Physiotherapy Dept
Woolmanhill
(01224) 555413

VOLUNTEER INFORMATION SHEET

Title: An investigation of the relationship between Anterior Knee Pain and lower limb biomechanics.

Introduction
I would like to invite you to take part in a study on knee pain. The study is aimed at finding out if pain at the front of the knee is affected by problems from the leg and foot. As part of the study I require information on " normal" measurements and volunteers who are free of knee pain will provide this measurements in order to determine whether the measurements of patients with knee pain is any different.

What will I have to do if I take part?
The study would take place at the Grampian Gait and Motion Analysis Centre, which is situated at Woodend Hospital, Aberdeen and would take up approximately 1 hour of your time. On your visit you would be asked to wear shorts and T-shirt (these can be provided if necessary).
You would be asked to fill out a short questionnaire and answer a few questions in a semi-structured interview.
Some body measurements would be taken (height, weight, and leg-length) and then some reflective markers would be attached to your skin with hypoallergenic double-sided tape at various points on your legs and feet. You would be asked to walk across the room at your normal pace whilst infra-red cameras detect the markers on your body and convert them into a 3-D image of you on a computer screen. This will be saved on the computer using a code and your name will not be used to identify the image. If, during this process you should feel any discomfort, the procedure can be stopped immediately.
The procedure would then be repeated but you would be asked to jog gently across the room.
The information collected would only be used by the researcher and the supervisor and would remain confidential at all times.

Are there any possible benefits?
This study will not benefit you directly. It is hoped that the information gained will help improve the clinicians knowledge of knee pain and help in the future development of treatment and prevention programmes for patients

Do I have to take part?
No, taking part is voluntary. If you would prefer not to take part you do not have to give a reason. This decision would not affect any future treatment you may require from the NHS. If you decide to take part but later change your mind, you can withdraw from the study at any time without giving a reason and this will not alter treatment.

What do I do now?
If you are interested or if you have any questions regarding the study please contact the researcher at the address below.

Thank you very much for considering to take part in this research. Any questions can be directed to:

Gillian Aitkenhead (Research Student)
Robert Gordon University
School of Health Sciences
Department of Physiotherapy
(01224) 555414

PATIENT INFORMATION SHEET

Title: An investigation of the relationship between Anterior Knee Pain and lower limb biomechanics.

Introduction

I would like to invite you to take part in a study aimed at finding out if pain at the front of the knee is affected by problems from the leg and foot and whether orthotics (a device which fits into your shoe) are an effective method of treatment.

What will I have to do if I take part?

The study would take place at the Grampian Gait and Motion Analysis Centre, which is situated at Woodend Hospital, Aberdeen and would take up approximately 1-2 hours of your time. On your visit you would be asked to wear shorts and T-shirt (these can be provided if necessary).
You would be asked to fill out a short questionnaire and answer a few questions in a semi-structured interview.
Some body measurements would be taken (height, weight, and leg-length) and then some reflective markers would be attached to your skin with hypoallergenic double-sided tape at various points on your legs and feet. You would be asked to walk across the room at your normal pace whilst infra-red cameras detect the markers on your body and convert them into a 3-D image of you on a computer screen. This will be saved on the computer using a code and your name will not be used to identify the image. If, during this process you should feel any discomfort, the procedure can be stopped immediately.
The procedure would then be repeated but you would be asked to jog gently across the room.
For the last part of the study, you would be asked to attend the MRI (Magnetic Resonance Imaging) centre, which is also situated at Woodend Hospital. Here you would have a scan of your lower legs with shoes on using a given orthotic and with shoes off using the upright MRI scanner. This machine differs from the conventional MRI, which you may be familiar with, in that it is not enclosed and you stand upright. You should not feel confined or claustrophobic. This process should only take 10-15 minutes and if at any time you wish to discontinue, the process can be stopped immediately.
The information collected would only be used by the researcher and the supervisor and would remain confidential at all times.

Are there any possible benefits?

This study will not benefit you directly. It is hoped that the information gained will help improve the clinicians knowledge of knee pain and help in the future development of treatment and prevention programmes for patients

Do I have to take part?
No, taking part is voluntary. If you would prefer not to take part you do not have to give a reason. This decision would not affect any future treatment you may require from the NHS. If you decide to take part but later change your mind, you can withdraw from the study at any time without giving a reason and this will not alter treatment.

What do I do now?
If you are interested or if you have any questions regarding the study please contact the researcher at the address below.

Thank you very much for considering to take part in this research. Any questions can be directed to:

Gillian Aitkenhead (Research Student)
Robert Gordon University
School of Health Sciences
Department of Physiotherapy
(01224) 555414

CLINICAL ASSESSMENT FORM

Date of gait analysis:

Name:

Gender:

Date of Birth:

Presenting complaint, if any:

Height (m):

Weight (kg):

Visual Observation: static posture (Tick box if appropriate)

Hips	Right	Left	Comments
Normal			
Flexed			
Extended			

Knees	Right	Left	Comments
Normal			
Flexed			
Extended			
Genu varum			
Genu valgum			
Genu recurvatum			
Patellar position high			
Patellar position normal			
Patellar position low			

Feet	Right	Left	Comments
STJ Normal			
Inverted			
Everted			
Navicular drop low			
Navicular drop normal			
Arch height high			
flat			
normal			
Base of gait narrow			
normal			
wide			
Angle of gait normal			
abducted			
adducted			

Presence of tibial torsion: Right yes/no Left yes/no

Use of assisted devices (including orthoses):

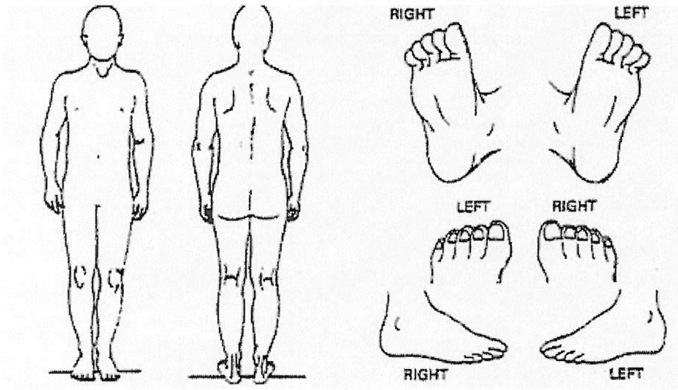
Any other observations:

Special tests done whilst weightbearing

	Right	Left	Comments
RCSP			
NCSP			
Navicular drop			
Q-angle			
Arch height			
Patella alignment			

Footwear/wear patterns:

Site of pain/lesions



Range of Movement: Supine

Hip	Right	Left	Comments
Flexion			
Extension			
Abduction			
Adduction			
External rotation (hip in neutral)			
Internal rotation (hip in neutral)			
External rotation (hip in 90 degrees)			
Internal rotation (hip in 90 degrees)			

Knee	Right	Left	Comments
Extension			
Flexion			
Q-angle			

Leg-length	Right	Left	Comments
Leg length			
Distance ASIS to ASIS			
Structural/functional			

Ankle	Right	Left	Comments
With knee flexed at 90 degrees Dorsiflexion			
Plantarflexion			
With knee extended Dorsiflexion			
Plantarflexion			

1st Ray	Right	Left	Comments
Dorsiflexion			
Plantarflexion			

1st MTJ	Right	Left	Comments
Dorsiflexion			
Plantarflexion			

Hamstring flexibility	Right	Left	Comments
Against gravity			
Gravity eliminated			

	Right	Left	Comments
Malleolar torsion			

Range of Movement: Prone

Quadriceps strength	Right	Left	Comments
Against gravity			
Gravity eliminated			

	Right	Left	Comments
Tibial torsion			

Feet	Right	Left	Comments
STJ Neutral			
Forefoot to Rearfoot			
Rearfoot inversion			
Rearfoot eversion			
Forefoot inversion			
Forefoot eversion			

CONSENT FORM

Name of volunteer:

Name of study: **An Investigation of the relationship between Anterior Knee Pain and power limb biomechanics**

Principal Investigator: **Gillian Aitkenhead;**

I have read the patient/volunteer information sheet and have had the opportunity to discuss the details with Gillian Aitkenhead and ask questions. She has explained the nature and purpose of the study, which I understand fully.

I have agreed to take part in the study as it has been outlined to me but I understand that I am completely free to withdraw from the study at any time I wish.

I understand that this is a small study which aims to investigate the effect of hip, knee, leg and foot measurements on knee pain and has been approved by the Grampian Research Ethics Committee and may be of no use to me personally. The Grampian Research Ethics Committee may wish to inspect the data collected at any time as part of its monitoring activities.

I hereby fully and freely consent to participate in the study, which has been fully explained to me.

Signature of patient/volunteer:

Date: 16 /12/ 03

I confirm that I gave an explanation to the patient/volunteer named above and purpose of the study to be undertaken.

Signature of Investigator: G. Aitkd.

Date: 16/12/03

!MKR#2

[input points]

LFHD	Left front head
RFHD	Right front head
LBHD	Left back head
RBHD	Right back head
C7	Back of neck
T10	Upper back
CLAV	Clavicle
STRN	Sternum
LBAK	Left back (optional alternative anti-sym)
RBAK	Right back (optional alternative anti-sym)
LSHO	Left shoulder
LUPA	Left upper arm (optional alternative anti-sym)
LELB	Left elbow
LFRA	Left forearm (optional alternative anti-sym)
LWRA	Left wrist bar thumb side (alternative pair 1)
LWRB	Left wrist bar pinkie side (alternative pair 1)
LWRI	Left wrist marker (alternative pair 2)
LWRE	Left wrist extender (alternative pair 2)
LFIN	Left finger
RSHO	Right shoulder
RUPA	Right upper arm (optional alternative anti-sym)
RELB	Right elbow
RFRA	Right forearm (optional alternative anti-sym)
RWRA	Right wrist bar thumb side (alternative pair 1)
RWRB	Right wrist bar pinkie side (alternative pair 1)
RWRI	Right wrist marker (alternative pair 2)
RWRE	Right wrist extender (alternative pair 2)
RFIN	Right finger
SACR	Sacral wand marker
LASI	Left ASIS
RASI	Right ASIS
LTHI	Left thigh wand marker
LKNE	Left knee
RTHI	Right thigh wand marker
RKNE	Right knee
LTIB	Left tibial wand marker
LANK	Left ankle
RTIB	Right tibial wand marker
RANK	Right ankle

```
LHEE    Left heel

LTOE    Left toe

RHEE    Right heel

RTOE    Right toe

LNAV
RNAV
LCALC1
LCALC2
LCALC3
RCALC1
RCALC2
RCALC3


LKAX    Left knee axis

LKD1    Left knee device 1

LKD2    Left knee device 2


RKAX    Right knee axis

RKD1    Right knee device 1

RKD2    Right knee device 2


LASI,RASI,SACR


LASI,LKNE,LTHI,LASI

RASI,RKNE,RTHI,RASI


LKNE,LANK,LTIB,LKNE

RKNE,RANK,RTIB,RKNE


LANK,LTOE,LHEE,LANK

RANK,RTOE,RHEE,RANK


LKAX,LKD1,LKD2,LKAX

RKAX,RKD1,RKD2,RKAX


{* Define Output Points - Virtual Navicular joints*}
```

```
[Output joint]

LNAVvirtual
RNAVvirtual

[Segment Axes]

PEL1
PEL2
PEL3
LFE0
LFE1
LFE2
LFE3
RFE0
RFE1
RFE2
RFE3
LTI0
LTI1
LTI2
LTI3
RTI0
RTI1
RTI2
RTI3
LFO0
LFO1
LFO2
LFO3
RFO0
RFO1
RFO2
RFO3

{*Usual VCM outputs*}

[Forces]

Point1

Force1

Point2

Force2

[Angles]

PelvisAngles

LHipAngles

RHipAngles

LKneeAngles

RKneeAngles

LAnkleAngles
```

RAnkleAngles

[Moments/Powers]

LHipMoment

RHipMoment

LKneeMoment

RKneeMoment

LAnkleMoment

RAnkleMoment

LPower

RPower

```
{*VICON BodyLanguage (tm)*}

{*VICON Clinical Manager Equivalent Model*}

{*For use only with BodyBuilder Version 3.1, or later*}


{* Show the force vectors *}

if EXIST( ForcePlate1 )
        Force1 = ForcePlate1(1)
        Moment1 = ForcePlate1(2)
        Centre1 = ForcePlate1(3)
        if ( ABS ( Force1 ) > 10 )
                Point1 = Centre1 + { -Moment1(2)/Force1(3),
Moment1(1)/Force1(3), -Centre1(3) }
        else
                Point1 = Centre1
        endif
        Force1 = Force1 + Point1
        OUTPUT ( Point1, Force1 )
endif

if EXIST( ForcePlate2 )
        Force2 = ForcePlate2(1)
        Moment2 = ForcePlate2(2)
        Centre2 = ForcePlate2(3)
        if ( ABS ( Force2 ) > 5 )
                Point2 = Centre2 + { -Moment2(2)/Force2(3),
Moment2(1)/Force2(3), -Centre2(3) }
        else
                Point2 = Centre2
        endif
        Force2 = Force2 + Point2
        OUTPUT ( Point2, Force2 )
endif
```

```
{*Pelvis*}
{*======*}

{*Hip Centres Estimation*}
{*If $UseTrocanter == 1 Then*}
{*      LAsisTrocDist = 10*$LAsisTrocDist*}
{*      RAsisTrocDist = 10*$RAsisTrocDist*}
{*Else*}

        LAsisTrocDist = 0.1288*$LLegLength*10-48.56
        RAsisTrocDist = 0.1288*$RLegLength*10-48.56

{*EndIf*}

{* Parameters used to determine position of Hip Joint Centres - From Kadaba and
Davis *}
C = (LAsisTrocDist+RAsisTrocDist)*0.0575-15.3
InterASISDist=DIST(LASI,RASI)
aa = InterASISDist/2
mm = $MarkerDiameter/2
COSBETA = cos(18)
SINBETA = sin(18)
COSTHETA = cos(28.4)
SINTHETA = sin(28.4)
COSTHETASINBETA = COSTHETA*SINBETA
COSTHETACOSBETA = COSTHETA*COSBETA

PELO=(LASI+RASI)/2
Pelvis = [PELO,LASI-RASI,PELO-SACR,yzx]

{* Note '-' signs: not present in earlier versions *}
LHJC = {C*COSTHETASINBETA - (LAsisTrocDist + mm) * COSBETA,
```

```
-C*SINTHETA + aa,
 -C*COSTHETACOSBETA - (LAsisTrocDist + mm) * SINBETA}*Pelvis

RHJC = {C*COSTHETASINBETA - (RAsisTrocDist + mm) * COSBETA,
        C*SINTHETA - aa,
        -C*COSTHETACOSBETA - (RAsisTrocDist + mm) * SINBETA}*Pelvis

{*OUTPUT(LHJC,RHJC)*}



{* Add point mass at pelvis origin *}
Pelvis = [Pelvis, Bodymass*0.142, {0,0,0}, {0,0,0}]



{*Femora*}
{*======*}

KneeOffset=($MarkerDiameter+$KneeWidth)/2

LKJC=CHORD(KneeOffset,LKNE,LHJC,LTHI)
RKJC=CHORD(KneeOffset,RKNE,RHJC,RTHI)

LFemur=[LKJC,LHJC-LKJC,LTHI-LKJC,zxy]
RFemur=[RKJC,RHJC-RKJC,RKJC-RTHI,zxy]

{*Femoral Rotation adjustment*}

LFemur=ROT(LFemur,3(LFemur),$LFemurRotation)
RFemur=ROT(RFemur,3(RFemur),$RFemurRotation)

LFemurLength=Average(DIST(LKJC,LHJC))
LFemurLength2=LFemurLength*LFemurLength
LFemurScale = LFemurLength/100
```

```
RFemurLength=Average(DIST(RKJC,RHJC))

RFemurLength2=RFemurLength*RFemurLength

RFemurScale = RFemurLength/100



{* Add the hierarchy, mass, and inertia *}

LFemur=[LFemur, Pelvis, LHJC, 0.1*BodyMass, LFemurLength*{0,0,0.567},
LFemurLength2*{0,0.292,0.292}*0.1*BodyMass ]

RFemur=[RFemur, Pelvis, RHJC, 0.1*BodyMass, RFemurLength*{0,0,0.567},
RFemurLength2*{0,0.292,0.292}*0.1*BodyMass ]



{*Tibiae*}

{*======*}


AnkleOffset=($MarkerDiameter+$AnkleWidth)/2


LAJC=CHORD(AnkleOffset,LANK,LKJC,LTIB)

RAJC=CHORD(AnkleOffset,RANK,RKJC,RTIB)


LTibia=[LAJC,LKJC-LAJC,LTIB-LAJC,zxy]

RTibia=[RAJC,RKJC-RAJC,RAJC-RTIB,zxy]


{*Tibial Rotation adjustment*}


LTibia=ROT(LTibia,3(LTibia),$LTibiaRotation)

RTibia=ROT(RTibia,3(RTibia),$RTibiaRotation)


LTibiaLength=Average(DIST(LAJC,LKJC))

LTibiaLength2=LTibiaLength*LTibiaLength

LTibiaScale = LTibiaLength/100

RTibiaLength=Average(DIST(RAJC,RKJC))

RTibiaLength2=RTibiaLength*RTibiaLength

RTibiaScale = RTibiaLength/100
```

```
{* Add the hierarchy, mass, and inertia *}

LTibia=[LTibia, LFemur, LKJC, 0.0465*BodyMass, LTibiaLength*{0,0,0.567},
LTibiaLength2*{0,0.279,0.279}*0.0465*BodyMass ]

RTibia=[RTibia, RFemur, RKJC, 0.0465*BodyMass, RTibiaLength*{0,0,0.567},
RTibiaLength2*{0,0.279,0.279}*0.0465*BodyMass ]


{*Feet*}

{*====*}


LFoot=[LTOE,LTOE-LAJC,LKJC-LAJC,xyz]

RFoot=[RTOE,RTOE-RAJC,RKJC-RAJC,xyz]


{*Apply Static Trial Offsets*}

LFoot=ROT(LFoot,2(LFoot),-$LStaticPlantarFlexion)

RFoot=ROT(RFoot,2(RFoot),-$RStaticPlantarFlexion)


LFootLength=Average(DIST(LAJC,LTOE))

LFootLength2=LFootLength*LFootLength

LFootScale = LFootLength/100

RFootLength=Average(DIST(RAJC,RTOE))

RFootLength2=RFootLength*RFootLength

RFootScale = RFootLength/100


{* Work out which way subject is walking *}
If 1(PEL0) > 1(SACR) then
  LAB = [{0,0,0}, {1,0,0},{0,0,1}, xyz]
else
  LAB = [{0,0,0}, {-1,0,0}, {0,0,1}, xyz]
endIf


{*Segment Axes*}

{*============*}
PEL1=PEL0+50*1(Pelvis)
```

```
PEL2=PEL0+50*2(Pelvis)
PEL3=PEL0+100*3(Pelvis)
LFE0=0(LFemur)
LFE1=LFE0+50*1(LFemur)
LFE2=LFE0+50*2(LFemur)
LFE3=LFE0+LFemurLength*3(LFemur)
RFE0=0(RFemur)
RFE1=RFE0+50*1(RFemur)
RFE2=RFE0+50*2(RFemur)
RFE3=RFE0+RFemurLength*3(RFemur)
LTI0=0(LTibia)
LTI1=LTI0+50*1(LTibia)
LTI2=LTI0+50*2(LTibia)
LTI3=LTI0+LTibiaLength*3(LTibia)
RTI0=0(RTibia)
RTI1=RTI0+50*1(RTibia)
RTI2=RTI0+50*2(RTibia)
RTI3=RTI0+RTibiaLength*3(RTibia)

LFO0=0(LFoot)
LFO1=LFO0+50*1(LFoot)
LFO2=LFO0+50*2(LFoot)
LFO3=LFO0+100*3(LFoot)
RFO0=0(RFoot)
RFO1=RFO0+50*1(RFoot)
RFO2=RFO0+50*2(RFoot)
RFO3=RFO0+100*3(RFoot)

{*Virtual Points Output*}
{*=====================*}
{*OUTPUT(PEL0,PEL1,PEL2,PEL3,LHJC,RHJC)*}
{*OUTPUT(LFE0,LFE1,LFE2,LFE3,RFE0,RFE1,RFE2,RFE3,LKJC,RKJC)*}
{*OUTPUT(LTI0,LTI1,LTI2,LTI3,RTI0,RTI1,RTI2,RTI3,LAJC,RAJC)*}
{*OUTPUT(LFO0,LFO1,LFO2,LFO3,RFO0,RFO1,RFO2,RFO3)*}

{*These virtual points are defined to make the segments visible in the Workspace
window, when a suitable MKR file is selected.*}



{*Yiannis*}

LNAVvirtual=LNAV/LFoot
RNAVvirtual=RNAV/RFoot

OUTPUT(LNAVvirtual,RNAVvirtual)



{* Add the hierarchy, mass, and inertia *}

LFoot=[LFoot, LTibia, LAJC, 0.0145*BodyMass, LFootLength*{0.5,0,0},
LFootLength2*{0,0.226,0.226}*0.0145*BodyMass]

RFoot=[RFoot, RTibia, RAJC, 0.0145*BodyMass, RFootLength*{0.5,0,0},
RFootLength2*{0,0.226,0.226}*0.0145*BodyMass]



{*Angles*}

{*======*}
```

```
global = [{0,0,0},{-1,0,0},{0,1,0},xzy]
PelvisAngles=<Pelvis,global,yxz>
PelvisAngles=<1(PelvisAngles),2(PelvisAngles),3(PelvisAngles)>
LHipAngles=<Pelvis,LFemur,yxz>
LHipAngles=<1(LHipAngles),2(LHipAngles),3(LHipAngles)>
RHipAngles=<Pelvis,RFemur,yxz>
RHipAngles=<1(RHipAngles),-2(RHipAngles),-3(RHipAngles)>
LKneeAngles=<LFemur,LTibia,yxz>
LKneeAngles=<-1(LKneeAngles),2(LKneeAngles),3(LKneeAngles)>
RKneeAngles=<RFemur,RTibia,yxz>
RKneeAngles=<-1(RKneeAngles),-2(RKneeAngles),-3(RKneeAngles)>
LAnkleAngles=<LTibia,LFoot,yxz>
LAnkleAngles=<1(LAnkleAngles),-2(LAnkleAngles),3(LAnkleAngles)>
RAnkleAngles=<RTibia,RFoot,yxz>
RAnkleAngles=<1(RAnkleAngles),-2(RAnkleAngles),-3(RAnkleAngles)>

{*OUTPUT(PelvisAngles)*}
{*OUTPUT(LHipAngles,RHipAngles,LKneeAngles,RKneeAngles,LAnkleAngles,RAnkleAngles
)*}


{*Kinetics*}
{*========*}

LHipMoment = 2(REACTION(LFemur))
RHipMoment = 2(REACTION(RFemur))

LKneeMoment = 2(REACTION(LTibia))
RKneeMoment = 2(REACTION(RTibia))

LAnkleMoment = 2(REACTION(LFoot))
RAnkleMoment = 2(REACTION(RFoot))

{*OUTPUT(LHipMoment,RHipMoment,LKneeMoment,RKneeMoment,LAnkleMoment,RAnkleMoment
)*}
```

```
LPowerHip =  POWER( Pelvis, LFemur )/BodyMass
LPowerKnee = POWER( LFemur, LTibia )/BodyMass
LPowerAnkle = POWER( LTibia, LFoot )/BodyMass
RPowerHip =  POWER( Pelvis, RFemur )/BodyMass
RPowerKnee = POWER( RFemur, RTibia )/BodyMass
RPowerAnkle = POWER( RTibia, RFoot )/BodyMass
{*OUTPUT(LPowerHip ,LPowerKnee,LPowerAnkle,RPowerHip,RPowerKnee,RPowerAnkle)*}
```

```
macro angleBetweenSticks(A,B,C,D,angle)
{* calculates the angle between the line joining A to B and the line joining C
to D Order is important! *}

AB = B - A
CD = D - C

angle#Temp = ACOS((AB(1)*CD(1) + AB(2)*CD(2) +
AB(3)*CD(3))/(DIST(A,B)*DIST(C,D)))

angle = <angle#Temp,0,0>
output(angle)

endmacro

C = {0,0,0}
D = {0,0,1}
angleBetweenSticks(LCALC1,LHEE,C,D,LInvAngle)
angleBetweenSticks(RCALC1,RHEE,C,D,RInvAngle)
```

CalcaneusSimpleAngles

```
macro angleBetweenSticks(A,B,C,D,angle)
{* calculates the angle between the line joining A to B and the line joining C
to D Order is important! *}

AB = B - A
CD = D - C

angle#Temp = ACOS((AB(1)*CD(1) + AB(2)*CD(2) +
AB(3)*CD(3))/(DIST(A,B)*DIST(C,D)))

angle = <angle#Temp,0,0>
output(angle)

endmacro

C = {0,0,0}
D = {0,0,1}
angleBetweenSticks(LCALC1,LHEE,LCALC2,LCALC3,LCalcAngle)
angleBetweenSticks(RCALC1,RHEE,RCALC2,RCALC3,RCalcAngle)
```

Inversion_eversion_3Dangles4

```
{* Inversion-eversion angles when foot defined by HEE, CALC and Virtual CALC
markers*}


{* Create virtual markers LVCALC, RVCALC *}

LVCALC = (LCALC1+LHEE+LANK)/3
OUTPUT (LVCALC)

RVCALC = (RCALC1+RHEE+RANK)/3
OUTPUT (RVCALC)


{* Create virtual markers for rear foot segment origins *}

LFootOrigin = (LANK+LHEE+LTOE)/3
OUTPUT (LFootOrigin)

RFootOrigin = (RANK+RHEE+RTOE)/3
OUTPUT (RFootOrigin)

LRearFootOrigin = (LCALC1+LHEE+LVCALC)/3
OUTPUT (LRearFootOrigin)

RRearFootOrigin = (RCALC1+RHEE+RVCALC)/3
OUTPUT (RRearFootOrigin)


{* Define segments *}

LRearfoot = [LRearFootOrigin, LCALC1-LVCALC, LHEE-LVCALC, xyz]
RRearfoot = [RRearFootOrigin, RCALC1-RVCALC, RHEE-RVCALC, xyz]
LFoot = [LFootOrigin, LANK-LHEE, LTOE-LHEE, xyz]
RFoot = [RFootOrigin, RANK-RHEE, RTOE-RHEE, xyz]


{* Calculate angles between segments *}

LInversion = <LRearfoot, LFoot, xyz>
OUTPUT (LInversion)

RInversion = <RRearfoot, RFoot, xyz>
OUTPUT (RInversion)
```

UNIVERSITY OF ABERDEEN **Research Volunteer safety form for MRI**

CONFIDENTIAL

To be completed by volunteer		
name:	date of birth:	weight:

To be completed by MRI staff	
MRI scan number:	Research supervisor / Dept:
Ethical Committee project number:	

The MRI scanner uses a very strong magnet and radio-waves to make images. This may be dangerous for some people. This questionnaire is designed to find out if it is safe for you to have an MRI scan.

Volunteers ***must*** complete this form before the examination. This information will be held securely within the MRI Department.

Please circle **Yes** or **No**

1) Do you have a heart pacemaker or implanted cardiac defibrillator?	1	Yes	No
2) Have you had any brain or heart surgery?	2	Yes	No
Do you have:			
3) an aneurysm clip?	3	Yes	No
4) an implanted nerve stimulator?	4	Yes	No
5) any artificial limbs or joints?	5	No	Yes
6) any metal plates or screws in your body?	6	Yes	No
7) a middle ear implant?	7	Yes	No
8) a hydrocephalus shunt (if yes, please give details over the page)?	8	Yes	No
9) any other implant, heart valve, coil, stent, catheter or artificial object in your body?	9	No	Yes
10) any metal fragments in your body (such as shrapnel or bullets)?	10	Yes	No
11) a false eye?	11	Yes	No
*12) dentures, dental plates or braces?	12	Yes	No
*13) a wig or hairpiece?	13	Yes	No
*14) a hearing aid?	14	No	Yes
15) a medication patch (nicotine / contraceptive)?	15	Yes	No
16) tattoos or tattooed eyeliner?	16	Yes	No
17) any body piercing jewellery?	17	Yes	No
18) any allergies?	18	No	Yes
*19) Have you had other operations (if yes, give details over the page)?	19	No	Yes
20) Have you ever had metal fragments in your eye?	20	Yes	No

Females Only

21) Are you or could you be pregnant?	21	Yes	No
22) Are you breast feeding?	22	Yes	No
23) Do you have a contraceptive coil?	23	No	Yes

If you have answered yes to any of the questions, please give additional information here ……………………………………………………..
…………………………………………………………………………………………..
…………………………………………………………………………………………..
…………………………………………………………………………………………..
…………………………………………………………………………………………..

IMPORTANT NOTES

- The University of Aberdeen cannot be held liable for any injury / death to the patient which arises as a direct result of any failure by the patient to disclose accurate and complete information on this form.

- **PLEASE DO NOT TAKE ANY METAL OBJECTS, COINS, BANK OR CREDIT CARDS, WATCHES ETC. INTO THE SCANNING ROOM. LEAVE THEM IN THE LOCKER PROVIDED IN THE CHANGING CUBICLE**

I confirm that I have read and understood the section entitled "Important Notes" above. I understand that I have the right to withdraw my consent to be imaged at any time without giving any explanation and that I will then be removed immediately from the MRI scanner. I confirm that the information given here is correct and complete to the best of my knowledge.

signed (volunteer): ……………………………………………. date: ……………

I am satisfied that the research volunteer has no contra-indications to exposure to the magnetic and electromagnetic fields associated with MRI, other than those explicitly allowed by specific Ethical Committee permission, associated with this research project.
Volunteers who have answered yes to any of the questions (except those marked with an asterisk) are not to be scanned unless specific Ethical Committee permission has been given.

signed (MR Authorised Person): ……………………….…. date: ………….

1 4 9 5

UNIVERSITY OF ABERDEEN **Patient safety form for MRI**

CONFIDENTIAL

To be completed by patient		
full name:	date of birth:	weight:

To be completed by MRI staff	
MRI scan number:	referring clinician / dept:
CHI number:	Unit number:

The MRI scanner uses a very strong magnet and radio-waves to make images. This may be dangerous for some people. This questionnaire is designed to find out if it is safe for you to have an MRI scan.

Patients ***must*** complete this form before the examination.
This information will be retained within your confidential medical record.

Please circle **Yes** or **No**

1) Do you have a heart pacemaker or implanted cardiac defibrillator?	1	Yes	No
2) Have you had any brain or heart surgery?	2	Yes	No
Do you have:			
3) an aneurysm clip?	3	Yes	No
4) an implanted nerve stimulator?	4	Yes	No
5) any artificial limbs or joints?	5	No	Yes
6) any metal plates or screws in your body?	6	Yes	No
7) a middle ear implant?	7	Yes	No
8) a hydrocephalus shunt (if yes, please give details over the page)?	8	Yes	No
9) any other implant, heart valve, coil, stent, catheter or artificial object in your body?	9	No	Yes
10) any metal fragments in your body (such as shrapnel or bullets)?	10	Yes	No
11) a false eye?	11	Yes	No
12) dentures, dental plates or braces?	12	Yes	No
13) a wig or hairpiece?	13	Yes	No
14) a hearing aid?	14	No	Yes
15) a medication patch (nicotine / contraceptive)?	15	Yes	No
16) tattoos or tattooed eyeliner?	16	Yes	No
17) any body piercing jewellery?	17	Yes	No
18) any allergies?	18	No	Yes
19) Have you had other operations (if yes, give details over the page)?	19	No	Yes
20) Have you ever had metal fragments in your eye?	20	Yes	No
Females Only			
21) Are you or could you be pregnant?	21	Yes	No
22) Are you breast feeding?	22	Yes	No
23) Do you have a contraceptive coil?	23	No	Yes

MRI screening form (patient) August 2009

If you have answered yes to any of the questions, please give additional information here ………………………………………………..
………………………………………………………………………………………..
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IMPORTANT NOTES

- The University of Aberdeen cannot be held liable for any injury / death to the patient which arises as a direct result of any failure by the patient to disclose accurate and complete information on this form.
- **PLEASE DO NOT TAKE ANY METAL OBJECTS, COINS, BANK OR CREDIT CARDS, WATCHES ETC. INTO THE SCANNING ROOM. LEAVE THEM IN THE LOCKER PROVIDED IN THE CHANGING CUBICLE**

I confirm that I have read and understood the section entitled "Important Notes" above. I further confirm that the information I have given on this form is correct and complete to the best of my knowledge.

signed (patient):	date:

or

signed (on behalf of patient):	date:
print name:	relationship to patient:

signed (MRI Authorised Person):	date: