“A Biomechanical Assessment of Gait Patterns and Risk of Associated Overuse Conditions among Mature Female Runners.”

Submitted by Kim Louise Lilley to the University of Exeter as a thesis for the degree of Doctor of Philosophy in Sport and Health Sciences In February 2012.

I certify that all material in this thesis which is not my own work has been identified and that no material has previously been submitted and approved for the award of a degree by this or any other University.

Signature: ……..

ABSTRACT

Due to a proliferation of health and social advantages, the popularity of running among the more mature members of the female population is expanding steadily. However, with both age and gender acting as possible risk factors, the incidence of running related injuries and associated conditions is high among this group. With the predominance of debilitating conditions such as knee joint osteoarthritis acting at the knee joint, knowledge of lower limb biomechanics during running will provide insight into possible risk factors and potential management strategies. Three biomechanical and one magnetic resonance imaging study focussed on the specific running gait of mature females and the effect of footwear on lower limb joint kinematics and loading. The biomechanical studies used synchronised ground reaction force and lower extremity kinematic data to provide three dimensional running data and knee moments for each female. The long term study objectives were to 1) determine whether the running gait of mature females could be a predisposing factor to injuries and conditions at the knee joint, and 2) determine if changes in footwear could modify biomechanical variables associated with the development of injuries and overuse conditions among this group.

In Study One, a direct comparison of mature and young female running gait was used to identify any biomechanical movement characteristics specific to the mature group that could predispose to injuries and debilitating conditions. It was found that rearfoot eversion, ankle dorsiflexion, knee internal rotation, and knee external adductor moment that are associated with increased loading of the lateral knee joint and possible medial knee joint osteoarthritis development, were significantly higher among the mature females compared to the younger group ($p<0.05$).

A common management strategy for running related conditions is the adaption of footwear. Therefore Study Two investigated the effect of a motion control running shoe on the running gait of young and mature females, with a specific focus on the variables associated with knee joint injury and osteoarthritis development. The results showed a motion control shoe to reduce certain biomechanical variables (rearfoot eversion and knee internal rotation) associated with mature female runners. However, one variable (knee external adductor moment) commonly associated with increased medial knee loading and osteoarthritis development, remained high among the mature females.

One specific method used to reduce the knee external adductor moment, is the implementation of a lateral wedge in running shoes. Therefore, Study Three assessed the singular effects of a medial wedge, a lateral wedge, and then the effect of an orthotic combining both interventions on the running gait of mature females. Results demonstrated non significant changes in any kinematic variable with the medial or lateral wedge, although the lateral wedge was shown to reduce the knee external adductor moment. The orthotic intervention however produced significant reductions in rearfoot eversion, knee internal rotation, and knee external adductor moment previously found to be high among mature female runners.

Although all mature females studied had previously been characterised as free from symptoms of knee injury or osteoarthritis, a final investigation was undertaken to assess the condition of the knee joint (Study Four). Magnetic resonance imaging scans of the knee were taken for ten of the mature females. Results indicated that eight out of the ten females had early stage osteoarthritis present, with an average 79% of features presenting on the medial side of the knee joint. Additionally, there was a strong positive correlation between knee

osteoarthritis and the knee external adductor moments measured in the ongoing biomechanical study (Study Three).

These studies have shown that the running gait of mature females is significantly different to that of younger female runners, and could predispose the mature group to injury and knee osteoarthritis development. The trends in kinematic adaption to a motion control shoe have shown promising results, and indicated the potential for footwear to reduce rearfoot eversion and knee internal rotation among mature female runners. However, a specific orthotic, incorporating both medial and lateral support has been found to reduce biomechanical features of gait associated with overloading at both the medial and lateral knee joint. The positive correlation between the knee adductor moment and signs of osteoarthritis for an asymptomatic population suggests that the knee adductor moment may be a useful predictive tool for identifying female runners at risk of osteoarthritis development.

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LIST OF DEFINITIONS.

TERMINOLOGY	DEFINITION
Angular Velocity	Rate of change of angular displacement.
Baby Boomers	Person born post World War II, between 1946 and 1964.
Calibration	Comparison of a measurement to a standard of known accuracy.
Gait	The pattern of movement of limbs during locomotion.
Ground Reaction Force	The force exerted by the ground on to a body.
Injury	Damage to soft tissue or bone of the musculoskeletal system.
Insole	See footbed. Can be altered to increase cushioning.
Joint Stiffness	Relationship between the deformation of a body and a given force.
Moment	A combination of the force applied to a segment, and the distance to the centre of rotation.
Motion Capture System	Combination of cameras and force plates used to assess human motion.
Muscle Strength	Propensity of a muscle to move a limb about a joint.
Orthotic	Orthopaedic device designed to support or alter the alignment of the limb or torso. Lateral wedge
Osteoarthritis	A multifactorial degenerative joint disease.
Overuse/Debilitating Condition	Degeneration of the bone or articular cartilage.
Plane of Movement	
Frontal	Longitudinal plane that divides the body into anterior and posterior sections.
Sagittal	Vertical plane that divides the body into medial and lateral sections.
Transverse	Horizontal plane that divides the body into superior and inferior sections.

Smoothing	Removal of high frequency noise from a data set.

Footwear Variables

Footbed	Manufacturer designed lining of shoe.
Orthotic Intervention	Full length lateral wedge with medial arch support.
Wedge	6mm (medial/lateral) wedge placed under footbed .

Variables of Gait

Abduction	Movement of a limb away from the midline of the body (frontal).
Adduction	Movement of a limb towards the midline of the body (frontal).
Eversion	Lateral tilt of the rearfoot on the oblique axis of the subtalar joint (frontal).
Extension	Movement of a joint causing an increased angle between two segments (sagittal).
Flexion	Movement of a joint causing a decreased angle between two segments (sagittal).
Inversion	Medial tilt of the rearfoot on the oblique axis of the subtalar joint (frontal).
Rotation	Rotation of a segment or joint about a rotation axis (transverse).

Abbreviations

KOOS	Knee Osteoarthritis Observation Survey
MRI	Magnetic Resonance Imaging

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Chapter 1. Introduction.

Exercise is defined as a form of physical activity that is planned, structured, repetitive, and purposeful in the sense that improvement or maintenance of one of more components of physical fitness is the key (Caspersen, Powell, & Christensen, 1985). Running is one form of physical activity that is commonly undertaken regularly, with the intent to improve cardiovascular fitness (Dunford & Doyle, 2011). In the spring of 2011, approximately 36,000 people took part in the London marathon, an increase from the 6,255 that completed the inaugural marathon in March 1981. Running as a recreational activity, appeals to all ages and genders across the world. Despite the fashion associated with recent gym based activities such as aerobics and dance related exercise classes, running still holds the interest of a large proportion of the population.

Regular participation in running has been proven to increase cardiovascular health, and decrease the risk of developing pathological conditions such as diabetes and psychological distress linked with depression (Thompson, Buchner, Pina, Balady, Williams et al., 2003; Knowler, Barrett-Connor, & Fowler, 2002; Pollock, 2001). Although popular among people of all ages, a particular group is becoming ever prevalent among the cohort of runners. In previous years, female runners comprised only 5 % of the population of runners, but with increasing popularity, the year 2008 showed this number to reach 50 % (Barrios, 2008). Furthermore, this increasing female cohort of runners has caused a re-characterisation of the sport, with fashion taking a headline in the industry almost to the exclusivity of males (Barrios, 2008).

With the baby boomers reaching retirement age and the mean age of the population continuing to increase, the popularity of running has expanded across all ages (Fukuchi, Eskofier, Duarte, & Ferber, 2011; Nigg, Fisher, & Ronsky, 1994). Due to its record of success both in weight loss, and stress relief, this particular sport is specifically becoming ever popular among the more mature generation of females. According to Barrios (2008), the pressure of creating balance between career, family, and health has caused running to be a "saving grace" for the more mature members of the female population; requiring minimal time, instruction, equipment, and economic expenditure (Barrios, 2008, pp.3).

Despite these benefits of running, according to Galloway (2005) the endorphins and "attitude boost" associated with this activity often drives the individual to ignore the early warning

signs of an injury or debilitating condition, until one erupts (Galloway, 2005; pp.201). The increasing number of people that take part in recreational and competitive physical activity has lead to a corresponding increase in the number of running related injuries and overuse conditions (Van Middlekoop, Kolkman, Van Ochten, Bierma-Zeinstra & Koes, 2008). In a systematic review of seventeen reports regarding lower extremity injuries and conditions, Van Gent (2007) reported the average incidence to range from 19.4 % to 79.3 %. These injuries and conditions present in a spectrum of severity, ranging from inflammation and pain to structural degeneration, with the knee joint the most common site for occurrence (Hootman, Macera, Ainsworth, Addy, Martin, & Blair, 2002). Among those sustained by runners, patellofemoral pain syndrome, shin splints, stress fractures, plantar fasciitis and osteoarthritis are among the most common, many appearing to be more common among women, and increase in prevalence with advancing age (Taunton et al., 2002).

Injuries and overuse conditions sustained by runners are often severe enough to cause a significant decrease or even cessation of training, and it has been reported that 12 to 44 % require medical attention (Brunet, 1990; Koplan et al., 1995). As one of the most common and disabling degenerative conditions, knee joint osteoarthritis has received a great level of attention within the literature. Knee joint osteoarthritis is considered to be twice as common among females when compared to males, and is known to substantially increase with age (Ballinger & Patchett, 2000). This suggests that the specific group of mature females is at a high risk of the condition.

Knee joint osteoarthritis is a degenerative disease of the articular cartilage, characterised by pain, inflammation and changes to the morphology of the joint that affects an abundance of people worldwide. Previously, radiographic techniques have been the primary method for identification of osteoarthritis; however more recently magnetic resonance imaging techniques have been considered to be superior (Cicuttini et al., 2004). In clinical settings, a range of surveys have also been validated to assess early signs of the condition (Roos and Toksvig-Larsen, 2003). Although the definitive causative factors and pathophysiology of knee joint osteoarthritis have not been confirmed, a clear association has been made between the biomechanical variable of knee external adductor moment and the initiation, development and progression of the condition (Hurwitz et al., 2000). As such, the use of techniques to assess biomechanical components of gait can be employed to investigate this condition among specific groups of runners.

Although the link between physical activity rates and the development of lower extremity injuries and conditions appears equivocal, it has been acknowledged that certain biomechanical variables during gait can predispose to the development of degenerative changes. During running, the movement of rearfoot eversion or subtalar joint pronation has received an increasing amount of interest among the biomechanical research community. It has been shown that a coupling action occurs between rearfoot eversion and internal rotation of the tibia and knee joint; both of which can predispose to injuries and debilitating conditions in the lower extremity (O'Connor & Hamill, 2004).

A fundamental method of limiting the rates of injury and development of overuse conditions is considered to be the application of suitable running footwear (Clarke et al., 1983). As such, the main item of equipment required for most runners is a suitable pair of running shoes. The concept of the athletic shoe was not considered until the American inventor Charles Goodyear patented the process of vulcanization of rubber. Since then, an abundance of designs and styles have become available, and highlighted by the vast number of advertisements in the sports magazines (for example, Runners World, Women's Running). One of the main concepts in running shoe design is the prevention and management of these sports injuries and debilitating conditions among runners. The idea behind these designs lies in the desire to ensure normal foot movement is achieved during the ground contact phase of running. Commonly, footwear attempts to control rearfoot eversion during running. Based on limited knowledge, an increasing number of people are under the impression that pronation is dangerous to runners, and many runners are purchasing trainers based on self diagnosis from uninformed internet sites. It is important to remain aware that subtalar joint pronation is a natural and essential motion required for normal biomechanics during gait. However, an excess of this motion has been linked with increased and uneven distribution of load on the structures of the lower extremity.

Although not unanimous in effectiveness, modifications to footwear have been described in the literature as an injury management strategy. Both wedging of the medial and lateral soles have been investigated in relation to injuries and conditions associated with subtalar joint pronation and the knee external adductor moment respectively (Clarke, Frederick & Hamill, 1983; Nester, Linden Bowker, 2003). However, these results have been disputed, and other researchers have suggested little effect of footwear modifications on the biomechanics during gait (Nigg et al., 1994).

It is suggested that alterations in footwear and orthotic design may have an influence on the running gait of mature female runners. Due to the high level of injuries and debilitating conditions such as knee joint osteoarthritis among this group, it is deemed important to investigate possible prevention methods. Currently, a proliferation of the research regarding footwear interventions have involved previously injured individuals, and therefore controversy exists as to the cause-consequence relationship between the biomechanics of running gait, footwear and the development of debilitating conditions at the knee joint.

The aim of this research thesis is to investigate gait patterns among mature female runners, to ascertain whether changes occur with age that could place this group at an increased risk of injuries and conditions. Certain footwear interventions will be investigated, to identify whether specific adaptations to the sole of the shoe may be suitable in the management of injuries and conditions for this particular group of runners.

Within this research project, an extensive review of the relevant literature is presented in Chapter 2, with specific focus on the topics that lead to the investigation of running injuries and conditions among mature female runners. The first study presented in Chapter 3 illustrates a direct comparison between the running gait of mature and young females, with a desire to further expand on the current theories regarding biomechanical changes with age, and the association of gait variables with injuries and debilitating conditions.

Chapter 4 presents an investigation of the effectiveness of motion control shoes to produce modifications in the running gait of mature and young females. This was performed to increase the knowledge regarding motion control shoes in potential injury prevention and management, and to assess the ability of footwear to alter biomechanics among mature female runners.

The third study is presented in Chapter 5, with a range of wedges and an intervention orthotic assessed. Each footwear condition was selected based on the previously illustrated pattern of running gait specific to mature female runners, with the intention of highlighting possible injury management techniques. An MRI study is presented in Chapter 6; performed to investigate the association between mature female runners and the signs of knee joint osteoarthritis development that formed the basis of this research project.

Chapter 2. Literature Review.

"We are under exercised as a nation. We look instead of play. We ride instead of walk. Our existence deprives us of the minimum of physical activity essential for healthy living"

John F. Kennedy (1917-1963).

2.1. Running: An Ability, A Sport, A Risk.

2.1.1. Human gait and the ability to run.

In today's society, a main challenge faced by the Department of Health is the growing epidemic of obesity, with the latest Health Survey for England (HSE) data showing nearly 1 in 4 adults are categorised obese (Department of Health, 2011). Similarly, the cost of obesity to the National Health Service is estimated at £4.2 billion, and is forecast to double by the year 2050 (Department of Health, 2011). This has lead to considerable media attention on the wider health risks of the sedentary routine led by many (Rose, Birch, & Kuisma, 2011). Habitual physical activity is therefore recognised as an important component of a healthy lifestyle, and running, as an accessible basic exercise, can be practised by almost any living human (Armstrong, 2007).

Today, millions of people run for the extensive benefits associated with the sport, including cardiovascular fitness and social expansion, and in the year 2008, over 34,000 people finished the London Marathon (Gifford, 2008). Regular participation in running causes a long term improvement in the circulatory system, with a decrease in resting heart rate and blood pressure, and an increase in stroke volume (volume of blood pumped with each heart beat) (Davis, Bull, Roscoe & Roscoe, 2000). Furthermore, regular participation in running improves muscle strength, increases cognitive functioning, and decreases the chance of psychological stress disorders (Weinberg & Gould, 2010).

Although the ability to run is an inborn quality among many, this skill requires the input of a variety of sources (Hawley, 2000). These include a talent for running, systematic coaching, adequate lifestyle habits, and most importantly an understanding of the body's systematic

functioning (Hawley, 2000). For running to be accomplished, a combination of actions from the body's systems is required, including the cardiovascular, digestive and endocrine systems. Additionally, a main factor that determines the ability of an individual to perform successfully is the human musculoskeletal system, which produces the process of running gait. Although the jointed lower limbs of most living humans are capable of a broad range of muscle-use and gait patterns, they generally prefer only two; walking and running (Srinivasan & Ruina, 2006). Running is accessible to most people, and with minimum equipment required, is the sport of choice for a large proportion of the population (Taunton, Ryan, Clement, McKenzie, Lloyd-Smith, & Zumbo, 2002).

2.1.2. Incidence of running related injuries and debilitating conditions.

According to Dishman and colleagues, an active lifestyle including regular running in comparison to sedentariness, is associated with a lower risk of premature death and developing chronic diseases such as coronary heart disease (Dishman, Washburn, Heath, 2004). However, it has been suggested that although a worldwide participation rate in running has been reported as 54.8%, the incidence rates of running related musculoskeletal injuries ranges from 19.4-79.3%. (Van Middlekoop, Kolkman, Van Ochten, Bierma-Zeinstra & Koes, 2008; Van Gent, Siem, Van Middlekoop, Van Os, Biernma-Zienstra & Koes, 2007). These injuries are commonly categorised into one of two groups; acute or chronic. Acute injuries occur from a single event where as chronic injuries tend to occur over time (Noakes and Granger, 1996). For the purpose of this thesis, the latter form is focussed upon, although both are initially considered.

Definitions of running injuries tend to differ between studies in the literature, and therefore make direct comparisons difficult (Van Gent et al., 2007). In general, running related injuries are most commonly soft-tissue injuries and stress fractures. In contrast, degeneration of the bone and articular cartilage is referred to as an over use, degenerative or debilitating condition (Van Gent et al., 2007). The cause of these running-related injuries and the development of running-related debilitating conditions have however both been subject to extensive research, with a variety of factors identified as risks (Chang, Shih, & Chen, 2011).

These include training errors, previous musculoskeletal injuries, incorrect shoe choice and malalignment (Lohman, Sackiriyas, & Swen, 2011). Training errors can be identified from an extensive knowledge of history, and commonly involve a sudden increase in mileage or intensity of training, or a change in environment, limiting the ability of the musculoskeletal system to adapt to the altered terrain (Migliorini, 2011). Similarly, an insufficient recovery period, or a lack of variation in training can often add to an individual's injury risk in the form of training errors (Hutson and Speed, 2011). Previous musculoskeletal injury has also been described as a risk factor, due to a compromise in the condition of the joint or muscle. Among football players, previous hamstring, groin or knee joint injury instigated a three times increase in the likelihood of sustaining an identical injury in the following season, and among long distance runners, a lower limb injury sustained previously was significantly correlated with its reoccurrence over time (Hagglund, Walden & Ekstrand, 2006; Middlekoop, Kolkman, Ochten, Bierma-Zeinstra, Koes, 2008).

Training errors and previous injury are examples of extrinsic and intrinsic risk factors respectively. Footwear and malalignment of the lower extremity are two more risk factors, similarly categorised as extrinsic and intrinsic factors respectively. However these two are linked, with appropriate footwear being frequently prescribed as a method of correcting biomechanical abnormalities during gait. The role of footwear in injury incidence and prevention is discussed in detail in a later chapter (4.2.7).

2.1.3. Location of running related injuries and debilitating conditions: the knee joint.

The dispersion of running related injuries is not even among the joints and muscles of the lower limb (Dishman et al., 2004). Figure 2.1 illustrates the percentage distribution of running related musculoskeletal conditions by lower limb body part among women. Injuries sustained to the upper body (32.6 %) are discounted.

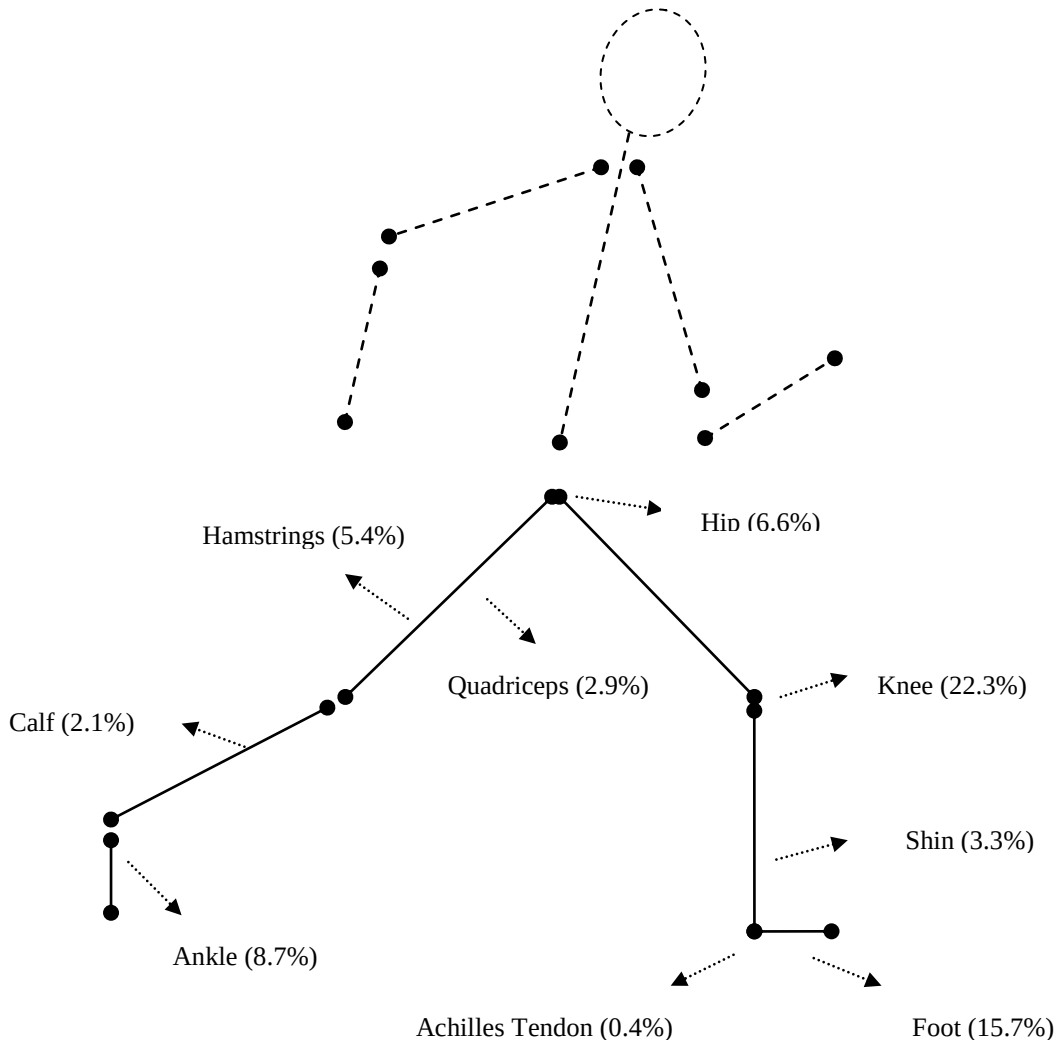


Figure 2.1. Percentage distribution of activity-related musculo-skeletal injuries isolated to the lower extremity. Adapted from Hootman et al., (2002).

According to Hootman and colleagues, the knee joint is the most common site for running related injuries, sustaining 22.3% of all injuries to the body, calculated from a sample of 260 active females (Hootman, Macera, Ainsworth, Addy, Martin, & Blair, 2002). This was also supported by Taunton and colleagues, who assessed running related injuries over a two year

period, and showed the knee joint to be the most common site for injury, accounting for 43.1 % of all injuries recorded (Taunton et al., 2002).

2.1.3.1. Anatomy of the knee joint.

The knee is the largest, most easily accessible, and one of the most complex joints in the human body (Logan & Rowe, 1994). Figure 2.2 illustrates a deep anterior view of the knee joint (Totora & Grabowski, 2003). Knowledge of the anatomy of the knee joint is paramount to understand the biomechanical function during gait, and mechanisms of knee injuries and conditions. Figure 2.2 illustrates the general basic anatomy of the joint, highlighting the major bones and ligaments. There are four major bones and two main joints at the knee; the patellofemoral joint and the tibiofemoral (Arokoski, Jurvelin, Vaatainen, & Helminen, 2000). The patellofemoral joint includes the intermediate articulation between the femur; the longest bone in the human body, and the patella; the largest sesamoid bone. At the tibiofemoral joint, the medial and lateral distal femoral condyles articulate with the corresponding plateaus of the tibia; the second longest bone in the human body (Logan & Rowe, 1994). The fibula is the fourth bone that articulates at the knee joint, positioned behind the lateral condyle of the tibia and acting as a site for the attachment of muscles and the lateral collateral ligament (Logan and Rowe, 1994).

Ligaments are present at the knee joint to restrict excessive knee excursion and maintain stability during stance and movement (Rybski, 2004). The two major ligaments are the cruciate ligaments, which join the femur and tibia within the articular capsule of the joint (Moore & Agur,1995). The anterior cruciate ligament (ACL) arises from the anterior intercondylar area of the tibia and extends superiorly, posteriorly and laterally to attach to the posterior lateral condyle of the femur. During knee extension, the ligament is taut, preventing posterior displacement of the femur on the tibia and hyperextension of the knee joint (Moore & Agur, 1995). Conversely, the posterior cruciate ligament (PCL) is the stronger of the two, and is the primary restraint to posterior movement of the tibia on the femur. It arises from the posterior intercondylar tibia, passing on the medial side of the ACL extending superiorly and anteriorly to attach to the medial femoral condyle (Rybski, 2004).

The two collateral ligaments, the medial collateral ligament and the lateral collateral ligament (MCL and LCL) resist movement in the valgus and varus directions respectively. When the knee is in the position of full extension, these two ligaments are taut, assisted by the posteromedial and lateral joint capsules, making extension the most stable position of the knee joint (Rybski, 2004). The iliotibial band (ITB) is a combination of the gluteus maximus and tensor fascia lateral tendons, and crosses the knee in the area of the lateral femoral condyle to insert to the Gerdy's tubercle (bony prominence just inferior and anterior to the lateral condyle of the tibia) (Behnke, 2006). This structure is also involved in controlling the flexion-extension movements of the knee joint.

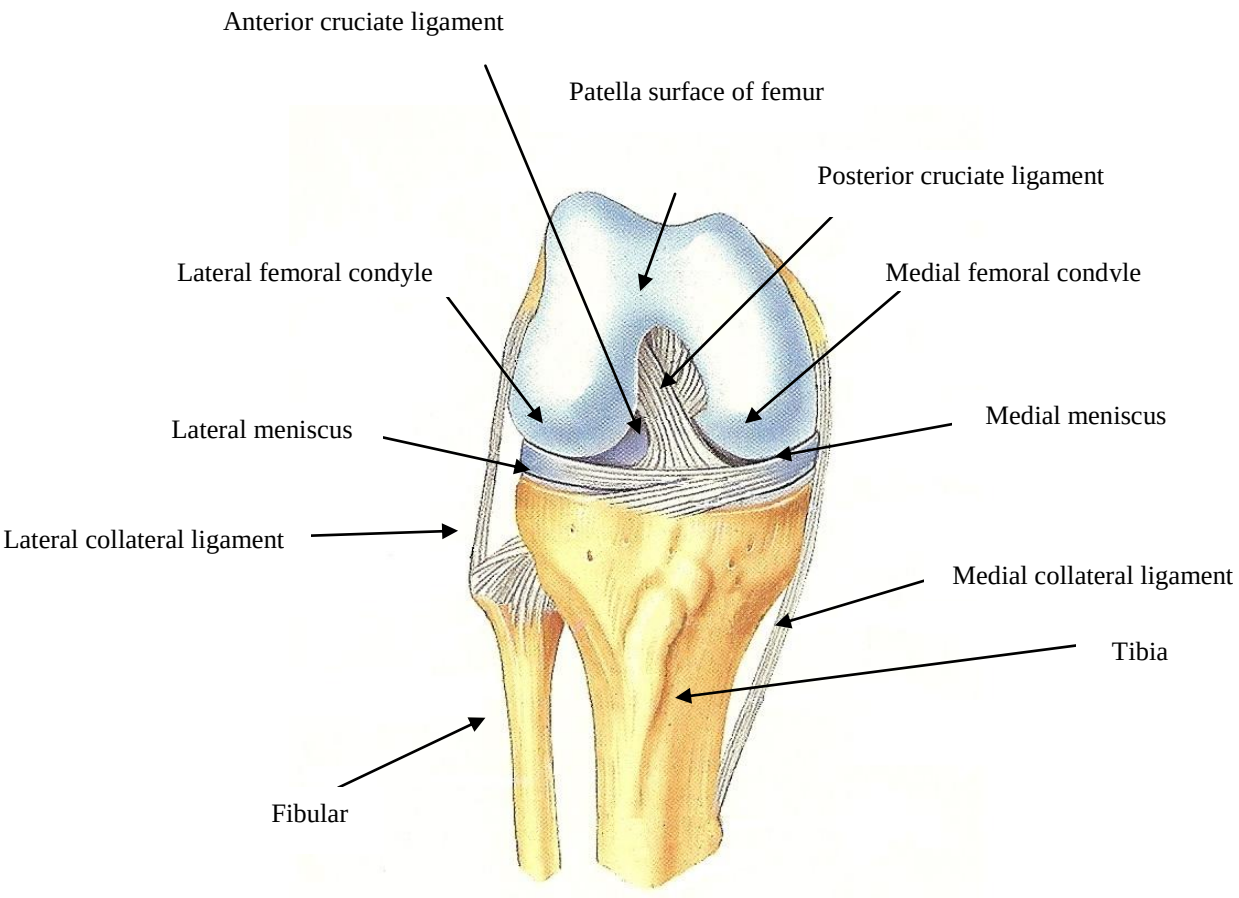


Figure 2.2. Anterior deep view of the knee joint. Source: Adapted from Totora and Grabowski, (2003).

As well as ligaments, a variety of tendons and muscles are involved in the function of the knee joint. Although approximately thirteen muscles act across the knee joint, most can be categorised into one of three main groups; the quadriceps, the hamstrings, and the calf muscles (Eaves, 2010). The tendons act to attach muscles to bone, and transmit the forces produced by the muscle groups across the joint. As shown in Figure 2.2, the knee joint also contains cartilage. The menisci are wedged-shaped cross sectional fibrocartilage structures that rest on the tibia at the anterior and posterior condyles (Norris, 2004). These structures receive blood flow from the genicular arteries and by adult hood, 10 – 25 % of the periphery of the meniscus is vascular. The menisci enable a better fit between the uneven shapes of the articulating bones of the knee joint, improving the bony correlation and shock absorption (Eaves, 2010).

Compared to fibrocartilage of the menisci, the articular cartilage is a soft tissue located at ends of bones; present to enable even distribution of forces and frictionless movement at the joint (Andriacchi, Mundermann, Smith, Alexander, Dyrby, & Koo, 2004). Articular cartilage is composed of chondrocytes within an organic matrix consisting of collagen fibrils (Fullick, 2000). The elastic characteristic of articular cartilage enables it to endure compressive forces and prevent wear of the articulating joint (Fullick, 2000). Detail regarding the internal structure and degradation of articular cartilage is described in a later section (2.1.4.1).

2.1.4. Conditions affecting the knee joint.

As described previously, injuries and conditions at the joints are commonly classified as either acute or chronic. Although acute injuries often occur and can result in debilitating consequences, this research thesis has focussed on the latter. A range of chronic conditions are known to affect the knee joint, including patella femoral pain syndrome, iliotibial band friction syndrome, meniscal injuries and osteoarthritis (Taunton et al., 2002). Various risk factors have been associated with many of these conditions, including gender, age, biomechanical alignment, and muscular functions. The influence of each factor is discussed in later sections.

2.1.4.1. Knee joint osteoarthritis.

As one of the most common conditions to present at the knee joint, osteoarthritis is given a particular focus within this thesis. Osteoarthritis is the degenerative disease of the joint; a process that involves the gradual loss of joint cartilage and changes in bone morphology (Totora & Grabowski, 2000). As the prevalence of knee osteoarthritis is expected to increase substantially with the aging of the baby boomers, it is important to understand the factors associated with knee osteoarthritis that may contribute to disability (Oatis, Wolff & Lennon, 2006). Osteoarthritis is commonly viewed as a disorder of cartilage and subchondral bone that results in a clinical syndrome of symptoms evolving from pathophysiologic changes within the joint (Dixon, Hinman, Creaby, Kemp, & Crossley, 2010).

Articular cartilage is a unique tissue with viscoelastic and compressive properties. Normal articular cartilage consists of a hydrated extracellular matrix containing non collagenous proteins, synthesised and maintained by a sparse population of specialised cells; the chondrocytes (Moskowitz, Altman, Hochberg, & Goldberg, 2007). The chondrocytes exist singularly or in groups, with spaces between them known as lacunae in the matrix. Although the weakest form of cartilage in the body, articular cartilage is flexible, allowing the bones to grow, and provides support at the articulating joints (Totora & Grabowski, 2003).

The extra cellular matrix of this cartilage is maintained in homeostatic balance by the catabolic and anabolic activity of the chondrocytes. Healthy articular cartilage in adults functions under a range of loading situations; although a change above or below the coping strategies of the cells may lead to cartilage degradation. Previous studies have supported this, suggesting that the loading environment of the joint is vital in maintaining activity of the chondrocytes. Griffin and colleagues highlighted the importance of loading for chondrocytes activity through in vivo assessment, and Arokoski and colleagues showed through an in-vitro epidemiological study, that moderate loading is vital in maintaining healthy cartilage (Griffin and Guilak, 2005; Arokoski, Jurvelin, Vaatainen, & Helminen, 2000). This theory is supported by Carter and colleagues who suggested that articular cartilage is healthiest when contact pressure and cartilage fluid pressure are high, although excessive and repetitive mechanical loading causes the initiation of articular cartilage degeneration (Carter, Beaupre, Wong, Smith, Andriacchi & Schurman, 2004). The effect of loading on cartilage form is illustrated in Figure 2.3.

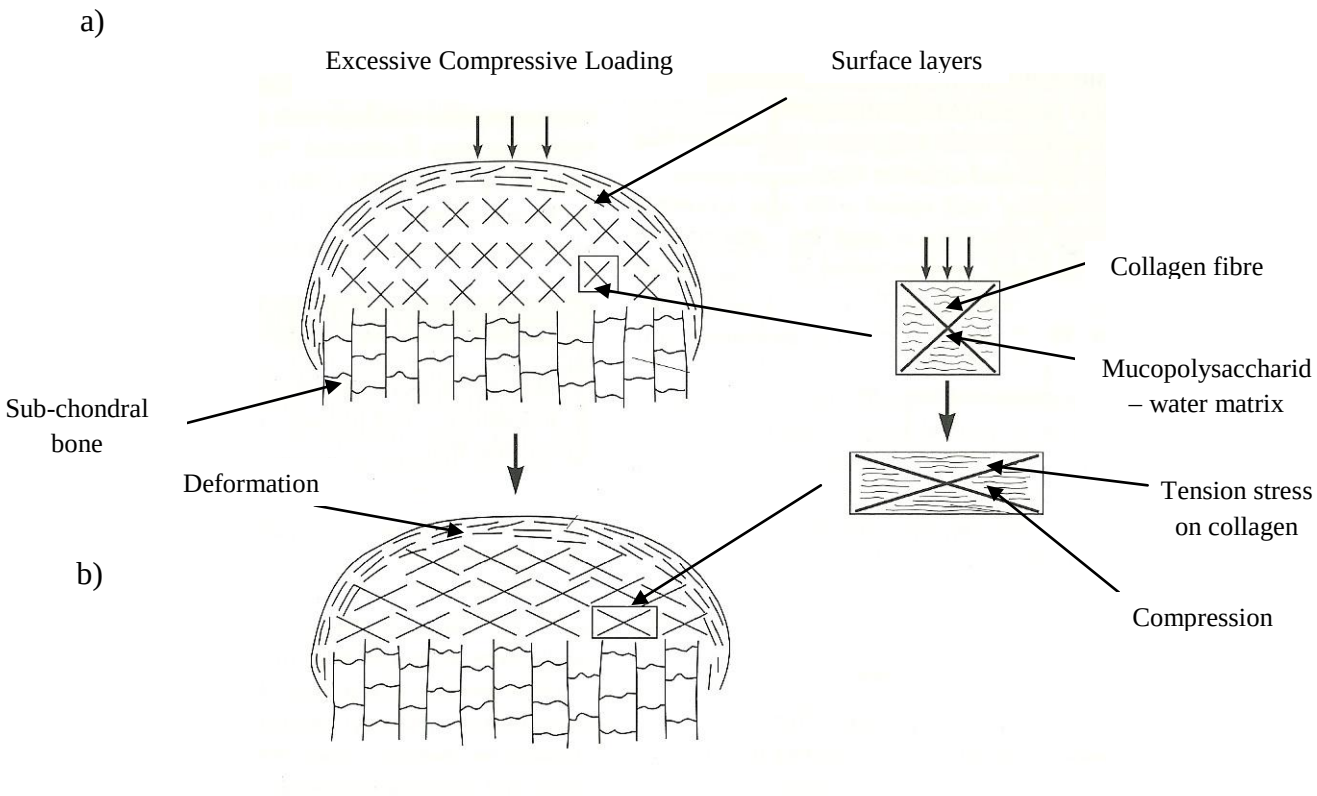


Figure 2.3. Effect of joint compression on articular cartilage; a) excessive compressive loading causes b) tension stress on the collagen fibres and matrix.

Source: adapted from Norris, (2004).

Although the pathogenesis of osteoarthritis remains unknown, it is commonly suggested that the condition is characterised by an imbalance between the anabolic and catabolic activities of cartilage cell population, with secondary inflammatory changes to the synovium and articular cartilage itself (Moskowitz, Altman, Hochberg, & Goldberg, 2007). Anabolic activities involve a construction of molecules from smaller units, whereby the enzymes cause

the chondrocytes to form cartilage. An excess of proteoglycan and collagen molecules then cause extra fluid to be diffused into the joint, diluting the chondrocytes, and causing the cartilage to be depleted. Conversely, catabolic activity involves an excess of destructive enzymes in the matrix, causing weakened collagen and lack of proteoglycans (Fox, Taylor, Yazdany & Brewer, 2006). Articular cartilage is formed from polymers, and as soon as degradation occurs, the bonds forming the polymer chains are destroyed, giving the condition it's predominantly irreversibly nature (Sabatini, Pastoureau, & De Ceuninck, 2004). Although cartilage repair and growth is possible, ostensibly cartilage is a metabolically inactive material, with limited blood supply and substances for repair. As such, the repair process requires blood cells to diffuse into the cartilage (Totora & Grabowski, 2003).

Although cartilage must be exposed to loading in order to maintain healthy operation, it has been suggested that excessive load, or an uneven distribution of load at the knee joint may instigate irreversible changes to the cartilage of the articulating joint (Roemhildt, Coughlin, Peura, Fleming, & Beynnon, 2006). During running, loads equalling between 1.5 and 5 times body weight are attenuated throughout the each leg during the stance phase of gait (Hreljac, 2004). As such, it could be considered that excessive impact forces attained during running could have a negative effect on the articular cartilage. The earliest studies involving animal models have shown mixed results, with running producing both minimal effects and harmful changes to articular cartilage.

According to Burton-Wurster and colleagues, compressive loads increasing from 0.025 to 1.2 MPa (mega pascal; unit of internal pressure) applied to canine articular cartilage over an 18 hour period caused a non linear inhibition of protein synthesis which would decrease enzyme concentration, limit the number of polymer chains formed and alter the assembly of the cartilage inner-structure (Burton-Wurster, Vernier-Singerm Farquhar, & Lust, 1993; Motaung, & Pieterse, 2011). However, it was also shown that intermittent loading of this cartilage partially prevented the increase in water content in the cartilage, suggesting that intermittent loading may help to maintain normal cartilage composition (Burton-Wurster et al., 1993). Roemhildt and colleagues supported the suggestion that high loads cause degenerative changes to articular cartilage in an investigation among rabbits (Roemhildt, et al., 2010). An in-vivo procedure illustrated a significant increase in permeability of articular cartilage with loads increasing from 0% increase in body weight to a 22% and 44% increase in body weight. Increased permeability represents decreased proteoglycan content in the

cartilage. This was also supported by the increase in cartilage thickness, which occurs from the swelling of the matrix, decrease in proteoglycan content, and disruption of the collagen fibres that resist the swelling pressure (Roemhildt et al., 2010). It was however noted here that these changes were site specific, and certain areas of the knee joint were more greatly affected when compared to others (Roemhildt et al., 2010).

Zhang and colleagues similarly showed changes to the articulating joint and cartilage structure, when loads of up to 17 MPa were applied to the joints of rabbits (Zhang, Vrahas, Baratta, & Rosler, 1999). In contrast, Borelli and colleagues also investigated the cartilage of rabbits, and showed single impacts of up to 55 MPa did not produce any significant change or signs of damage to the tissue, including enzyme activity, changes in proteoclycan concentration, or collagen breakdown (Borrelli, Zhu, Burns, Scandell, & Silva, 2004).

The equivocal results regarding loading effects on articular cartilage have contributed to the lack of definition for the pathogenesis of osteoarthritis. However it is reasonable to assume that abnormal loads are a likely primary component of the pathogenic mechanism of knee joint osteoarthritis. As well as magnitude of load, joint malalignment modifies the inter-segmental compressive loads across the articulating joint, and has been shown as a primary risk factor for development of osteoarthritis (Felson, Anderson, Naimark, Walker, & Meenan, 1988). As such, biomechanical factors can create loading situations that may increase load on specific areas of the knee joint and articular cartilage. Commonly, it is suggested that factors such as age, gender, strength and reduced joint range of motion (stiffness) can contribute to the susceptibility of the condition, and although many may act in combination, identifying the contribution from individual factors is helpful in isolating specific populations that are more prone to the condition. Each factor is discussed in detail in the following sub chapters.

2.2. Mature Female Runners.

Advances in health and medical services has lead to overall longer life expectancy (Kim, Lochart, & Yoon, 2005) and a greater proportion of mature people taking part in regular physical activity (Henriksson & Hirschfeld, 2005). However, a selection of studies has considered age as a risk factor for the development of injuries or joint conditions such as osteoarthritis (Fukuchi, Eskofier, Duarte, & Ferber, 2011). According to Bus (2003), the greater incidence of injuries and conditions among this specific active group may be due to age related changes in musculo-skeletal properties such as strength or joint stiffness, or biomechanical movement patterns during gait.

2.2.1. Age related changes of the musculo-skeletal system.

Increasing age is not merely a passage of time, but a manifestation of biological events that occur over a period of time (Robergs & Ketyian, 2003). Physiologically, certain changes occur with increasing age, including a reduction in maximal oxygen uptake during exercise (VO_2max) of 8% to 10% per decade after the age of 30 years (Robergs & Ketyian, 2003). This is supports McArdle, Katch and Katch (2001) who showed average VO_2max scores as 31 to 38.9 $mL.kg^{-1}.min^{-1}$ for females aged at or below 29 years, and 21 to 32.9 $mL.kg^{-1}.min^{-1}$ for those aged 60 to 69 years. As VO_2max is defined as the maximal rate at which the body can absorb oxygen during exercise (Robergs & Ketyian, 2003), this suggests that changes occur to the physiological system; the ability of the muscles to use available oxygen and the combined ability of the cardiovascular and pulmonary system to transport oxygen in the blood may reduce with age.

Advancing age is also known to instigate changes in the structure of bone, ligaments and tendons. As age increases, longitudinal trabaculae bone becomes thinner, and transverse trabaculae bone is resorbed (Nordin & Frankel, 1989). These result in a marked decrease in cancellous bone and a thinning of the cortical bone, in turn increasing the relative brittleness and limiting the energy storing capacity. These changes commonly occur first at the head of the femur, leading primarily to changes in hip function (Nordin & Frankel, 1989). A decline

in bone mineral density has been shown among men and women aged 55 years and over, with significantly greater decline in bone mineral density of the femoral neck among women compared to males (Burger, Van Daele, Algra, Ouwelan, Grobbeem Hofman et al., 1994). This suggestion of gender related differences in the reduction of bone mineral density with age was also supported by Lim and colleagues, who stated that body composition changes with age differ between males and females, altering the changes to bone structure and function (Lim, Joung, Shin, Lee, Kim, Shin, Kim, Lim & Cho, 2004).

As well as age related changes to the bone structure and function, certain alterations occur in the composition of tendons and ligaments within the body. According to Nordin and Frankel (1989), up to the age of 20 years, the number and quality of cross links within and between the collagen molecules of the tendons and ligaments steadily increases, producing an increase in the tensile strength of the structure. However, after maturation there is a plateau in the structure of the tendon or ligament, followed by a slight decline in the mechanical properties and the ability to withstand deformation during loading (Nordin & Frankel, 1989). This notion is supported by Baratz and colleagues who suggested that although biochemical analyses have not definitively shown dramatic age related changes in tendon or ligament matrix composition, the collagen and water composition tends to decline as the collagen cross links decrease in number and mechanical adaptation to load is compromised (Baratz, Watson, & Imbriglia, 1999).

Due to the invasive nature of monitoring ligament and tendon internal structure over time, the majority of research that has lead to these conclusions is based on studies involving animals. Woo and colleagues investigated changes in the medial collateral ligament among rabbits over a period of time, and showed a reduction in tensile strength of the structure at 18 months compared with that measured at 12 months (Woo, Ohland & Weiss, 1990). Similarly, Smith and colleagues showed tendons in older horses were less able to respond to exercise training, compared to tendons in younger horses that demonstrated an increase in fibril diameter and distribution (Smith, Birch, Patterson-Kane, Firth, Williams, Cherdchutham, Weeren, & Goodship, 1999). Noyes and Grood (1976) also showed the anterior cruciate ligaments from middle aged and elderly humans (48-86 years) failed by avulsion or detachment of the bone under the ligament insertion site more frequently than the same ligaments from younger humans (16-26 years).

2.2.2. Age related changes to gait.

Although many of the more mature members of the population are taking part in recreational and competitive running (Fukuchi et al., 2011), previous studies analysing gait changes with age have predominantly involved males during walking activities (Nigg, Fisher & Ronsky, 1994). Similarly, many have enabled a self selected walking pace to be analysed, and as gait speed and subsequent joint and segment velocities influence kinematic variables, these may not show a true reflection of age related changes in biomechanics (Hagerman & Blanke, 1986).

Grabiner and colleagues investigated differences in gait patterns of two groups of walkers (72.13 ± 3.96 years; 25.06 ± 4.02 years) and showed the only age related difference in gait to be an increased variability in stride width among the older group (Grabiner, Biswas & Grabiner, 2001). An earlier study investigating walking gait changes among young (23.9 ± 3.6 years) and mature (66.9 ± 7.6 years) women however showed the elderly group to display significantly smaller stride length, reduced ankle range of motion, and reduced pelvic obliquity (axial tilt) than the younger group (Hagerman, & Blanke, 1986). It was however noted that the speed selected by the younger group was significantly faster than that of the elderly group and no screening occurred to negate the possibility of a pathological condition that may have affected gait. Marigold and Patla (2008) showed a similar decrease in gait speed among a group of more mature walkers (74.1 ± 7.2 years) compared to a younger group (26.1 ± 5.2 years), and a shorter stride length. This concept was highlighted by Nigg and colleagues who suggested that decrease in walking speed commonly seen among the more mature members is associated with a reduction in step length as opposed to cadence (number of steps per unit of time) (Nigg et al., 1994). Additionally, Marigold and co-workers showed increased medial to lateral acceleration of trunk centre of mass, and increased trunk roll among the more mature group, suggesting decreased stability compared to the younger group (Marigold & Patla, 2008).

Although not extensive in the research literature, changes in joint angles with age have been shown among the more mature members of the population. As suggested previously, Hagerman and Blanke (1986) showed significant reductions in ankle range of motion among elderly women during walking gait, which supports the earlier work of Murray and colleagues, who showed decreased heel rise and ankle extension among males at the end of

stance phase of gait (Murray, Kory, & Clarkson, 1969). Nigg and colleagues showed significant changes in gait pattern with increasing age, and interpreted these as a consequence of changes in muscle strength in individuals over the age of 50 years (Nigg et al., 1994). In one investigation, a reduction was demonstrated in ankle range of motion, and an increase in the inversion-eversion passive range of motion, combined with increased foot abduction among the mature group (Nigg et al., 1994).

In a later study of age related changes in running gait, Fukuchi and Duarte (2008) illustrated mature runners (67 – 73 years) to show significant decreases in knee range of motion in the sagittal plane and significant increase in foot abduction (toe-out) angle compared to a group of younger runners aged 26 – 36 years. This was later supported by Fukuchi and colleagues who showed a similar discrepancy in knee flexion range of motion between male runners aged between 69 ±2 years and 31 (±6) years (Fukuchi et al., 2011). The combination of research suggests that gait related changes occur with age in walking and running. However increased research is required to assess the changes during running, with speed of gait controlled to eliminate possible variation in self selected running speeds between age groups.

2.2.3. Gender differences in gait and injury risk.

In addition to age, gender is considered to have an effect on an individual's risk of developing a running related injury or condition. According to Taunton and colleagues, when compared to males, females are twice as likely to sustain certain running related injuries and debilitating conditions (Taunton et al., 2002). It has been postulated that this could be due to differences in musculoskeletal structure and function during gait (Ferber, McClay, Davis, & Williams, 2003). Obvious anatomical and less obvious physiological differences exist between males and females, both of which often favour males in the performance of sports and physical activity that requires strength and size (Armstrong, 2007).

Many of the differences that exist between males and females can be attributed to the average increased size of physique of the male in comparison to the female physique. Often, when variables are corrected for or scaled to size, these changes are markedly reduced or abolished (McKean et al., 2007). Although changes between males and females occur during puberty or

maturation, in adulthood, differences in bone mass are suggested to occur as a result of the physically smaller female skeleton. According to Brukner and Khan (2006), a physiological difference exists in body fat composition, where average values for males and females are 14 % and 26 % respectively. This is combined with relatively less muscle mass among females. This difference is due to the levels of the female hormone oestrogen, which causes fat to be deposited at certain anatomical sites such as the hips and thighs (Martin & Gerstung, 2005).

Under the influence of the male hormone testosterone, males have a wider shoulder girdle, narrower pelvis and longer limbs compared to females. One of the main biomechanical differences between males and females is therefore the quadriceps angle (Q angle). This angle is created by the wide pelvis relative to shorter femur length in the female body, and is commonly associated with a varus (abducted) position at the hips, anteversion (forward rotation) of the femoral head and valgus (adduction) at the knees (see Figure 2.4) (Norris, 2004; Arendt, 1994). The Q angle is created between the force line of the quadriceps muscle, and the line of pull of the patella tendon (Comfort, & Abrahamson, 2010). Compared to males, females have a wide pelvis in relation to shoulder width and limb length, which, as a positive advantage, can lead to a slightly lower centre of mass, and potentially improved balance and stability (Brukner & Khan, 2006). It is however important to note that this increased Q angle is not universal among females, and it has been suggested that although some females have greater valgus angles of the thigh, this is uncommon among female athletes. According to Atwater (1990), generalisations that all females are inferior to males in running due to a wider pelvis and increased femoral obliquity have frequently been cited with little scientific proof.

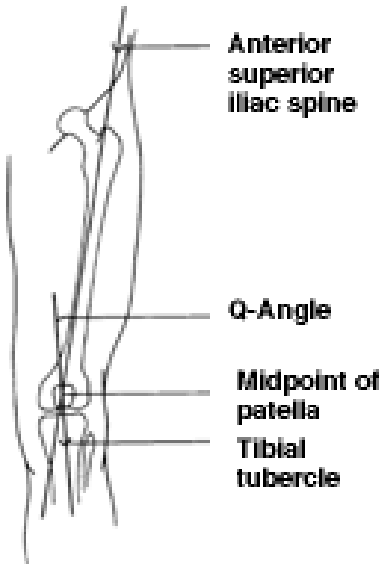


Figure 2.4. Illustration of the Q angle; calculated between the line of quadriceps and the vertical line through the mid point of the patella. Source: Adapted from Schumacher, (2012).

Among individuals with high Q angles, calculating this variable has been described as a method of assessing potential knee problems, due to an excess pull on the anterior cruciate ligament of the knee joint, and could be one explanation for the increased incidence of running related injuries among females (Comfort & Abrahamson, 2010). The results from an early study using cadaver knee joints supported this theory. Huberti & Hayes (1984) used pressure sensitive film to assess the pressure within the knee joint, at varying Q angles, and showed a 10 degree increase in Q angle to increase the peak pressure in the knee joints (Huberti & Hayes, 1984). Similarly, another in vitro study showed an increase in Q angle applied to cadaver knee joints increased the lateral patellofemoral contact pressure and could increase risk of patella dislocation (Mizuno, Kurnagai, Mattessich, Elias, Ramrattan, Cosgarea, Chao, 2001). According to Messier and colleagues, increased Q angles among females could play a partial role in the increased incidence of patellofemoral conditions that women experience (Messier, Davis, Curl, Lowery, & Pack, 1991). Conversely, Caylor and colleagues investigated the relationship between Q angle at maximum extension (0° flexion) and anterior knee pain syndrome and showed no association between higher Q angles and the

incidence of the condition (Caylor, Fites, & Worrell, 1993). Similarly, Bierdert and Warnke (2001) showed no association between Q angle at full extension and the position of the patella. It is noted however that all participants in this investigation were suffering with patella femoral pain syndrome at the time of investigation (Bierdet & Warnke, 2001). Additionally, these latter researchers suggested that the varying methods of measuring the Q angle, including both static and dynamic measurements, and the position of the knee joint could be the cause for the equivocal results presented in the literature.

In addition to the aforementioned biomechanical disparities, males and females have also been shown to display different patterns of gait. An increase in stride length has been shown for males when compared to females (Kerrigan, Todd, & Della Croce, 1998), although this has been shown to disappear when the data is normalised for height (McKean et al., 2007). According to Roislien and colleagues, when compared to males, women have been shown to walk with increased plantarflexion at toe off and initial swing, increased knee extension during stance and increased knee flexion during swing phase. Additionally, women were shown to walk with increased knee valgus (adduction) angles and increased hip internal rotation angle throughout the entire gait cycle (Roislien, Skare, Gustavsen, Broch, Rennie, & Opheim, 2009). It is suggested that these latter two movements are due to the previously described larger quadriceps angle inherent to females. Ferber and colleagues also showed a greater hip adduction angle among female runners when compared to males, and although not measured, postulated that this could be due to the gendered structural difference of hip width to femoral length ratio (Q angle) (Ferber, McClay Davis, Williams, 2003).

Cho and colleagues also showed differences between males and females during walking, and illustrated increased anterior pelvic tilt and increased hip flexion among the female participants (Cho, Park & Kwon, 2004). It was suggested that the increased hip flexion could be due to a learned behaviour of walking in high heels, or a decrease in the muscular strength of the abductor muscles. Additionally, females were shown to walk with decreased ankle plantarflexor moment, increased knee valgus angle, and increased hip adduction and abduction moment (Cho et al., 2004). The relatively smaller feet inherent to females could reduce the length of the lever arm of the sagittal moment at the ankle, reducing the plantarflexor moment and limiting the energy at the push off phase of gait (Cho et al., 2004). Increased hip internal rotation was also demonstrated among the female group, similar to that produced in the earlier study by Roislien and collagues (Cho et al., 2004; Roislein et al.,

2009). This difference in hip internal rotation was also demonstrated by a group of participants during running trials (Ferber et al., 2003). As well as increased hip internal rotation, the females in this investigation showed significantly greater knee valgus, combining to suggest an increased Q angle. According to Cho et al (2004), this biomechanical variable could be due to the anatomically shorter length of the female femur. Malinzak and colleagues investigated gender differences among participants during running, and illustrated similar frontal plane motion, but an average of 11° increased knee valgus among females when compared to males. Additionally, less knee flexion range of motion was shown for the female group (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001).

Although differences exist between studies, it has been shown that an increase in hip internal rotation and adduction angle appears to be common among females when compared to their male counterparts. This is commonly combined with an increase in knee joint valgus and often a reduction in sagittal plane range of motion at the knee joint. It is suggested that the increase knee valgus could contribute to various running related injuries and conditions that affect female runners (Chumanov, Wall-Scheffler, & Heiderscheit, 2008).

2.2.4. Influence of hormones and menopause.

The main female hormone, produced by the ovaries, is oestrogen. This hormone, along with follicle stimulating hormone (FSH) and luteinising hormone (LH) act to regulate the menstrual cycle (Totora & Grabowski, 2003). It has been proposed that an increase in risk of injury or development of running related debilitating conditions may be contributed to by hormonal fluctuations associated with the female menstrual cycle (Clark, Bartold, & Bryant, 2010). The majority of ligaments within the human body possess oestrogen and relaxin receptors, suggesting that these structures, and the potential for damage, may be influenced by the stages of the menstrual cycle (Clark et al., 2010). Oestrogen has also been linked with cardiovascular health, with increased levels of the hormone associated with the activation of endothelial nitric oxide synthesase, leading to vasodilation and reduced risk of arterial blockage (Mendelsohn, 2002).

Oestrogen is thought to passively diffuse into the cells in the body, and as such has been shown to have an effect on bone structure and remodelling. Throughout a life time, millions of small packets of bone are constantly remodelled through the balance between the amount of bone formed and resorbed. Bone resorption describes the process by which osteoclasts break down bone and release minerals, causing a transfer of calcium from bone fluid to blood (Krassas & Papadopoulou, 2001). After damage or wear on a bone, the major action of oestrogen may be to decrease this resorption of bone as opposed to aiding the formation of new bone (Krassas & Papadopoulou, 2001). According to the results from Liu and Howard, a treatment of oestrogen causes osteoclasts to lose attachment to the bone surface, undergo degenerative changes and limit the ability to resorb bone (Liu & Howard, 1991). As such it could be proposed that oestrogen protects bone from degenerative changes over time.

During the menopause, oestrogen levels decrease dramatically, due to the cessation of menstruation and the menstrual cycle. As such, bone loss or reduced bone remodelling during or after the menopause is common in women over the age of 50, and can be considered an additional risk factor for the development of degenerative conditions at the joints (Kanis, 1996). This was supported by Felson and colleagues, who suggested that the high incidence of osteoarthritis among post menopausal women infers that oestrogen deficiency plays a role in causing disease (Felson et al., 2000). The association between hormone replacement therapy (oestrogen tablets) and reduced incidence of radiographic osteoarthritis further supports this theory (Zhang, McAlindon, Hannan, Chaisson, Klein, & Wilson, 1998).

2.3. Analysis of Human Gait.

2.3.1. Human movement.

The production of coordinated human movement is achieved through a physiological process, combining actions from the nervous, muscular and skeletal systems of the human body (Davis, Bull, Rosco & Roscoe, 2000). According to Vaughan and colleagues, the cyclic movement of human gait is achieved through the following sequence of events:

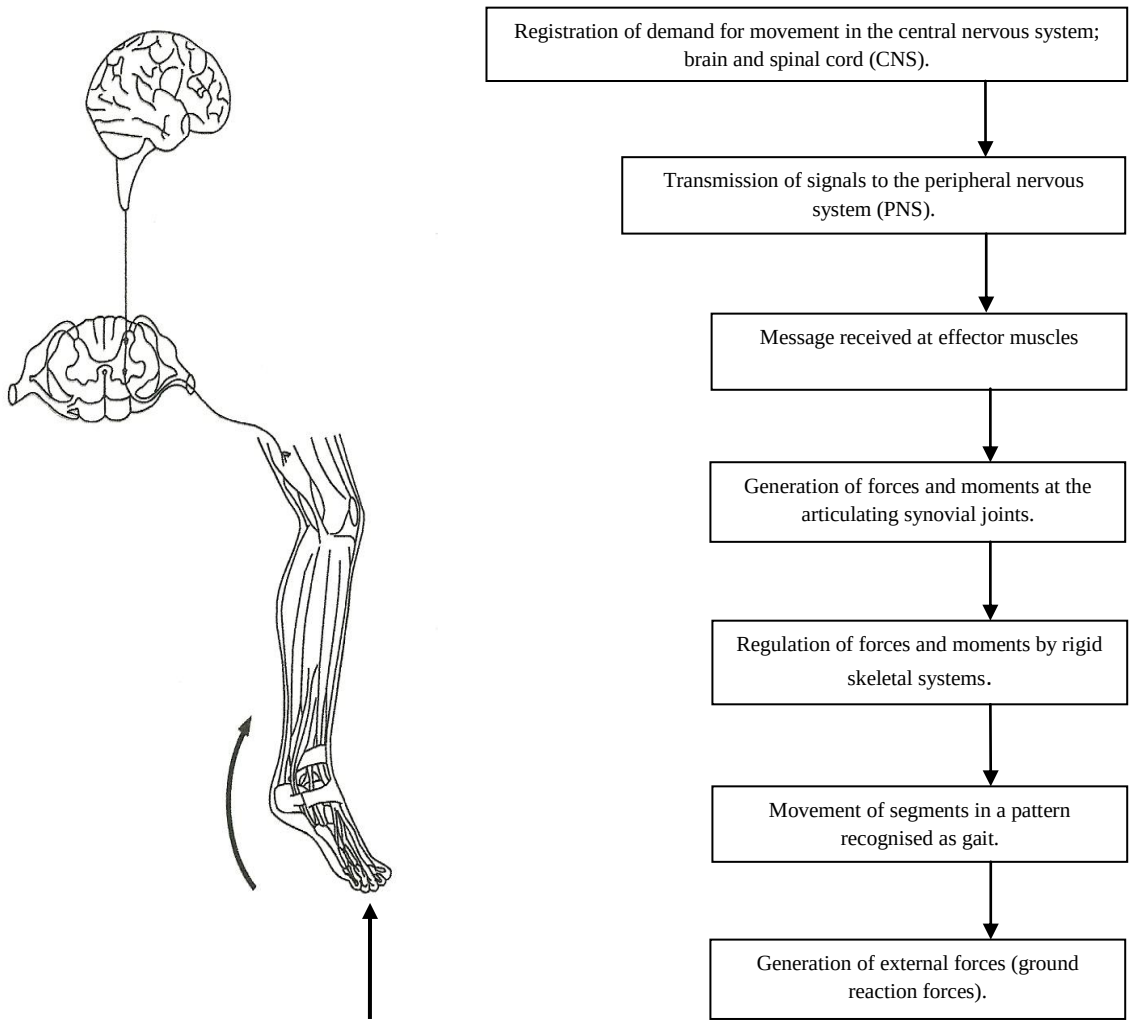


Figure 2.5. Process of movement produced through transmission of signal from the central nervous system. Source: Adapted from Vaughn, Davis & O'Connor, (1999).

Muscles can only contract and produce movement when a nerve ending is stimulated from an outgoing impulse sent from the central nervous system. Each contractile system is organised into a number of sections, each one controlled by a single motor neurone (Davis et al., 2000). When activated, the muscles develop tension, which generates forces and moments at the articulating joints, and produce movement at the rigid skeletal links (Vaughan, Davis, & O'Connor, 1999). This sequence of events is required to enable the cyclic movements of walking and running; motion commonly classified as gait (Vaughan et al., 1999).

2.3.2. Three dimensional gait analysis.

Three dimensional assessment of gait has been widely accepted as capable of quantifying gait and assisting in surgical decision making (Dorocjak & Cuddeford, 1995). The three dimensional aspect enables joint motion to be identified in the three planes of movement and, according to Chambers and Sutherland (2002), enables identification of rotational movements in the transverse plane, that are commonly mistaken with coronal or sagittal plane abnormalities, or subject to perspective error when using two dimensional methods. Early systems of kinematic analysis commonly captured data at slow sampling frequencies of 50 Hz or 60 Hz. This rate indicates how often the instrument records a measurement, and is therefore often selected based on the speed of the movement being captured (McGinnis, 2005). As such, the terms motion analysis and gait analysis now tend to refer to specialised hardware and software for collecting, processing and analysing kinematic and dynamic parameters of movement, and commonly involve systems including Peak Motus or Vicon (Peak Vicon, Oxford Metrics Co. Ltd, UK) (Hirsch, 2000). These systems permit the acquisition of coordinates of a series of reflective markers attached to anatomical landmarks, the spatial trajectories of whom are intended to be measured (Giannini et al., 1994).

2.3.3. Smoothing and filtering kinematic data.

Most methods employed for the analysis of movement introduce an amount of noise, or high frequency fluctuations from a signal (Bogert, 1996). These can often be removed after identification of the source. As such, a smoothing or filtering operation is performed on the coordinates of each maker to remove the noise. Within biomechanics, these two terms are often used interchangeably, and data smoothing techniques are commonly equivalent to low-pass filters. Filtering generally describes the removal of high frequency components from a signal, as noise has been suggested to be of a higher frequency than the frequency of the true signal (Winter, Sidwall, & Hobson, 1974). This method has been proved successful in removing noise from movement data; in a comparison between angular acceleration data acquired from accelerometers and synchronised video data, the digital filter effectively reduced signal noise and provided an accurate representation of the kinematic parameters (Pezzack, Norman & Winter, 1977). It is however noted that care is required when using the filtering technique, to ensure the process does not change the movement data itself (Richards, 2008). Smoothing techniques commonly employs the use of piece wise polynomials to determine a line of best fit curve through a set of data points (Wood & Jennings, 1979). A smoothing spline is a series of polynomials joined together at the nodes. This process is commonly used in conjunction with motion analysis software, and involves a smoothing parameter that controls the smoothness and closeness of fit (Richards, 2008). This is therefore used to represent the smooth nature of human movement and reject the random noise in the digitised coordinates (Bartlett, 2007). A noise factor is employed within the motion analysis system (Vicon, Oxford) which defines the maximum distance a marker can deviate from a parabola with respect to time, and is scaled relative to the calibration units to ensure the magnitude is dependent on the size of the capture volume. Noise factor set too high or low may cause the system to create false paths or delayed path assignments respectively.

Different smoothing techniques have been reported in previous studies for the reduction of noise from the raw displacement data. In a direct comparison of the effectiveness of a cubic spline, quintic spline, and digital filter, Vaughan (1982) illustrated the quintic spine to be superior to the other methods. This was supported by Challis and Kerwin (1988) who suggested that the quintic spline smoothing techniques is most accurate in producing the first and second derivatives, supporting its use in calculations of acceleration (second derivative).

2.3.4. Errors in kinematic assessment.

Despite the technological advancement of three dimensional gait analysis, as well as noise within the data, collection of biomechanical kinematic data is still often impeded by errors. As discussed previously, within two dimensional analyses, parallax and perspective error occur as subjects move away from the optical axis of the camera and out of the intended capture volume (Kirtley, 2006). However, although superior in ability to view tri-planar motion, when performing three dimensional motion analyses, error is encountered in many other areas.

All measurement that involves biological phenomena is subject to error. One of the main forms of error in the assessment of human gait is the assumption of rigid body segments, and subsequent marker placement. The mathematical model of the lower limb on which the markers are placed is a series of rigid linked bodies which move in space with six degrees of freedom; three linear translation, and three orthogonal angular rotations (Gianni et al., 1998). As such, within this form of analysis, it is assumed that soft tissue dynamics are ignored, all segments are rigid and the placement of markers on the anatomical landmarks represents the kinematics of the segment during gait (Nigg and Herzog, 1994). However the positions of these markers are subject to two forms of error; absolute and relative. Intrinsically linked, these two forms refer to the error produced in the movement of two markers that define rigid segments, and that produced from the movement of a marker relative to the anatomical landmark it is representing (Nigg and Herzog, 1994). Both forms of error can therefore be expected due to the movement of the soft tissue on which the marker is places. According to the results from a study comparing skin mounted markers, and markers attached to a plaster of paris mould of the limb assuming rigid segments, Capazzo (1991) illustrated a difference in movement of the underlying bone compared to the skin marker. During walking, displacement between markers was shown as 10 and 7 mm for the markers on the greater trochanter and femoral condyle respectively (Capazzo, 1991). Reinschmidt and colleagues also supported this theory, and showed that when comparing with bone pins secured to the underlying bone, external skin markers produced oscillations in rotation values calculated at the knee joint (Reinschmidt, Van Bogert, Lundberg, Nigg, Stacoff, & Stano, 1997). However it was noted that the external markers did reflect the knee extension – flexion movements and the tibiocalcaneal rotation movements. Lundberg (1996) has also stated that marker error is

effected by the specific marker assessed; the marker placed on the anterior shank where the skin is relatively tightly held across the bone will represent the bony movement to a superior level when compared to a marker on the knee joint (Lundberg, 1996). This is supported by Kirtley (2006) who suggested that the hip joint marker is the marker most prone to absolute error, due to the level of soft tissue around the pelvis and hips, especially among women. However although methods such as securing bone pins directly to the underlying bone may reduce these errors, it is noted that this method is not ethically acceptable for routine motion analysis (Richards, 2008).

Forms of absolute error are commonly encountered with regard to the clothing worn by participants, and whether markers are affixed to footwear or directly to the skin of the foot-ankle complex; common factors in gait analysis. In one comparison between the markers placed on skin and those on lycra to represent the bony landmark of the anterior superior iliac spine, Hazelwood and colleagues illustrated significantly more movement of the skin mounted marker, suggesting lycra to be preferable in error limitation (Hazelwood, Hillman, Lawson,, & Robb, 1997). Analysis of rearfoot motion has also been shown to be affected by marker placement, and according to Stacoff and colleagues, comparisons of results between studies is difficult due to overestimation of joint movement with both skin and show mounted markers compared to markers affixed to the bone (Stacoff, Nigg, Reinschmidt, Bogert, & Lundberg, 2000). Morio and colleagues have discussed this error, and in one investigation, chose to look at the effects of sandals rather than shoes, due to the proposed difficulty of monitoring foot movement inside a shoe (Morio, Lake, Guillaume, Baly, 2009). Similarly, Wolf and colleagues attempted to overcome the marker error by attaching markers at cut out points in the shoe, to enable the markers to be attached directly to the skin (Wolf, Simon, Patrikas, Schuster, Armbrust, & Doderlein, 2008). Shultz and Jenkyn, (2011) further investigated this method, and investigated the optimal diameter of the hole in the shoe to enable sufficient maker movement but little change in footwear properties. It is therefore suggested that due to the specificity of the size and shape of the hole in the shoe, this method may not be applicable to studies involving large groups of participants required to wear the same footwear, due to individual variation in foot shape.

2.3.4.1. Reliability, repeatability and accuracy.

When performing gait analysis, a level of error exists in any measurement taken; which directly affects the accuracy of the system. The data can only be manipulated to define parameters that are clinically significant if there is sufficient reliability, repeatability and accuracy of the system in question (Giannini et al., 1994). Reliability determines the capacity of the system to give a faithful representation of movement in different sessions, and can be expressed as the minimum standard deviation acquired from the measurements (Gianni et al., 1994). Conversely, repeatability refers to the stability of the measurement over time; the correlations between reproductions of the same movement from the same subject in the same session (Gianni et al., 1994). Accuracy is a measure of the difference between the true value and the observed or measured value produced from the system (Ehara, Fujimoto, Tanaka, & Yamamoto, 1995). According to Allard and colleagues, the root mean square error (RMSE) is a statistic used to illustrate the difference between the observed and known values, and can therefore be used to represent and compare the accuracy of a system (Allard, Blanchi, & Aissaoui, 1995). An assessment of the accuracy of the system used in this research project is provided in the appendix, Appendix A. The root mean square error provides a standard deviation and is calculated using the formula below.

$$RMSE = \sqrt{\sum(X-\alpha)^2/(n-1)}$$

Equation 2.1

The root mean square error is calculated where X is the known or true value, α is the measured value and n refers to the number of observations. A measurement of a point location is taken, and the arithmetic error is calculated as the difference between the measured and true value. This is squared, and the square of differences are combined. The square root of this sum is then divided by the number of observations minus one, to produce a root mean square error or accuracy that can be compared between systems.

2.4. Kinematic Movement Data.

2.4.1. Assessment of joint centres and angles.

The purpose of gait analysis is to track the motion of the body segments during a walking or running movement (Whittle, 2003). The model of the limb on which the methodology of this process is based is that of a series of rigid linked segments that move in space in six degrees of freedom (Giannini, Catani, Benedetti & Leardini., 1994). This refers to the three linear translations; medial-lateral, vertically, and anterior-posterior motion, and the three rotational movements in the sagittal, coronal, and transverse planes (Richards, 2008). It is through the tracking of the coordinates of each segment that the orientation of one with respect to another is determined, therefore defining the joint angles produced (Gianni et al., 1994). The joint centre locations of the hip, knee and ankle are determined through the placement of markers within the mathematical marker set model. The location of joint centres is required to be calculated in an accurate yet repeatable manor, as expressed by Della Croce and colleagues who stated that error in anatomical definition is more detrimental than errors caused by skin or soft tissue movement (Della Croce, Capozzo & Kerrigan, 1999). In general, the joint centres are determined using palpable anatomical landmarks to define the medial-lateral axis of the joint (Richards, 2008). As such, the ankle joint centre is located at the midpoint between the lateral malleous on the dome of the talus and the joint line between the tibia and the talus. This is calculated through the position of two markers, placed on the lateral and medial malleolus. The knee joint centre is located at half of the knee diameter at the joint line between the femur and the tibia at the lateral femoral condyle, as the axis of rotation of the knee passes through this landmark (Kirtely, 2006). Similarly, calliper measurement enables knee diameter calculation. Accuracy of joint centre determination is vital, and according to Holden and Stanthorpe (1998), a marker displacement of only 10 mm can have a great effect on the joint angles calculated at the knee.

At the hip joint, the movement between the femur and pelvis is assumed to be a pure rotation about the joint centre (Cereatti, Donati, Camomilla, Margheritini, Capozzo, 2009). However determining the hip joint centre is difficult, due to the lack of bony landmarks, and the excess soft tissue common among females leading to increased absolute marker error. This was supported by Cereatti and colleagues who showed error in hip joint centre location to be

greater among participants that had greater soft tissue artefacts (Cereatti et al., 2009). During gait analysis, the hip joint centre is commonly identified through a functional method, whereby the femur is moved passively or actively, to locate the middle of the head of the femur within the sphere formed by the acetabulum (Della Croce, Leardini, Chiari, & Capozzo, 2005).

A joint angle is that which lies between two connecting segments that share a common point. Two dimensional angles are generally determined counter clockwise from the first to the second segment. Conversely, three dimensional angles are calculated as the internal angle between two segments, and range from 0 to 180 degrees. Within this thesis, joint angles are determined through using the joint coordinate system (Figure 2.6), whereby one coordinate axis is used from each local coordinate system of the two segments that constitute the joint. This method employs simple trigonometry to calculate joint angles (Equation 2.2 a-c).

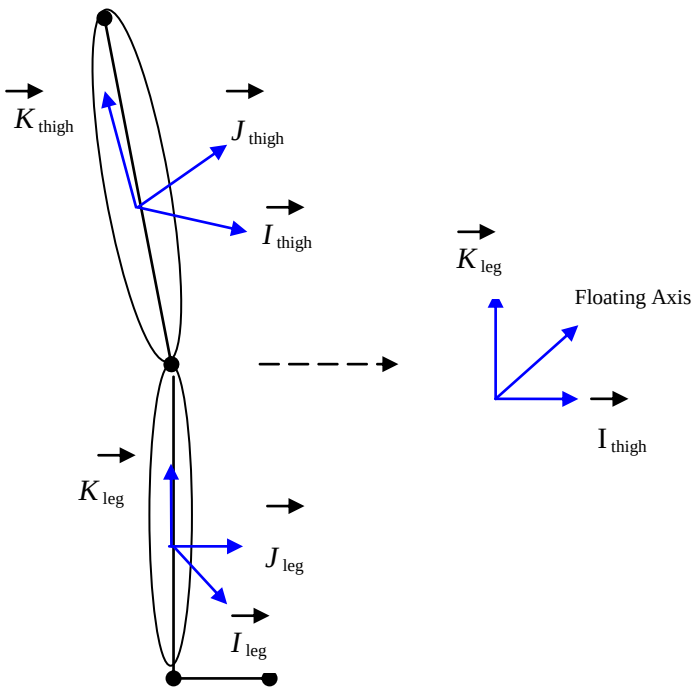


Figure 2.6. Joint coordinate system for the knee. Source: Adapted from (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2004, p. 52).

a) flex-ext = $90 - \cos^{-1}(K_{proximal} \cdot FA)$

b) abd-add = $-90 - \cos^{-1}(K_{distal} \cdot I_{proximal})$

c) int-ext = $90 - \cos^{-1}(I_{distal} \cdot FA)$

Equation 2.2a-c

Figure 2.6 illustrates a joint coordinate system for the knee, whereby the longitudinal axis is the vertical axis of the distal segment and the lateral axis is the medial-lateral axis of the proximal segment. The third axis is a floating axis (FA) that is the cross product of the longitudinal and lateral axis and acts in a perpendicular direction (Robertson et al., 2004). Description of the anatomy and joint movements of the ankle and knee joints are detailed in the following sections.

2.4.2. Ankle motion: Dorsiflexion/plantarflexion.

The complete movement of the foot as an object about the tibia is referred to as ankle joint motion (Richards, 2008). The ankle is a weight-bearing joint, the movements of which strongly determine the pattern of gait (Norris, 2004). This joint is referred to as the talocrural joint, as it is formed by the bones of the leg and a single tarsal bone, the talus (Jenkins, 1998). Overall, it consists of the trochlear surface of the talus, and the distal ends of the tibia and fibula. Figure 2.7 illustrates the ankle joint and the motions involved.

The relationship among these three bones is maintained by musculotendinous structures, ligaments and a fibrous capsule (Alter, 2004). There are three articulating surfaces within the joint; the upper, medial and lateral. The former is the main articulation of the joint, cylindrical in shape, and formed by the tibia and talus (Whittle, 2003). This joint is

essentially a hinge joint, with movement almost entirely limited to flexion and extension in the sagittal plane (Norris, 2004). The two main movements at the talocrural joint are dorsiflexion and plantar flexion.

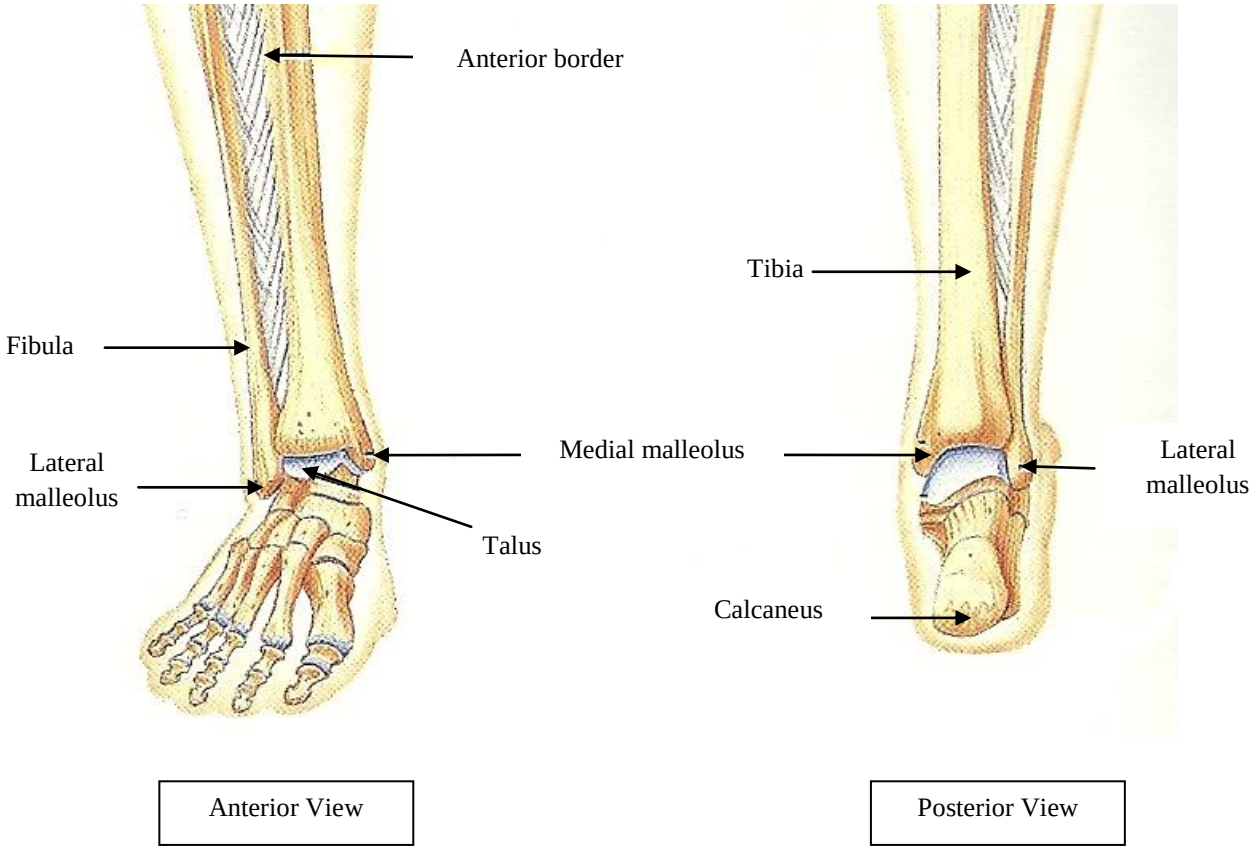


Figure 2.7. Right limb illustrating the bones at the ankle joint. Source: Adapted from Totora & Grabowski (2003) p.236.

Plantar flexion of the ankle involves extension of the foot (Alter, 2004). This motion causes the narrow posterior part of the trochlear surface of the talus to move into the broader tibiofibular mortice (Norris, 2004). This motion is primarily produced by the actions of the

gastrocnemius and soleus muscles on the calcaneus (Martini & Nath, 2008). The flexor hallucis longus, flexor digitorum longus, and the tibialis posterior muscles in the calf also play an important role in assisting plantar flexion, with their tendons sent posteriorly around the medial malleolus (Jenkins, 1998). Finally, the peroneus longus and brevis contained in the lateral compartment can also produce plantar flexion of the ankle (Dutton, 2005).

In contrast to the motion of plantar flexion, dorsiflexion of the ankle forces the broad anterior section of the talus into the narrower mortice between the tibia and fibula (Norris, 2004). Simply, this causes the foot to be pulled towards the shin, and as such the tibialis anterior is therefore the most important muscle involved (Jenkins, 1998). Similar to plantar flexion however, there are additional muscles that can aid the motion, and in the case of paralysis of the former muscle, dorsiflexion of the ankle is produced through the actions of the extensor digitorum longus and associated peroneus tertius (Jenkins, 1998). The major ligaments of the ankle are those between the tibia and fibula, their primary function being to prevent distance forming between these two bones (Whittle, 2003). During dorsiflexion, the joint moves into its most stable closed pack position and the tibiofibular ligaments are stressed (Norris, 2004).

Both plantar flexion and dorsiflexion are movements essential to normal running gait. During a heel strike running gait, the importance of this overall movement of the foot to the tibia is based on its role in shock absorption at heel impact and its function in the propulsive stage immediately prior to toe off (Richards, 2008). Peak dorsiflexion of the ankle occurs as the tibia moves over the ankle during the latter stage of midstance, and maximum plantar flexion is reached as the heel lifts at the end of the stance phase (Richards, 2008).

Ankle joint motion in the sagittal plane is calculated using the position of the two segments; the foot and the tibia. As described previously, the segment angles are calculated using trigonometry (Robertson et al., 2004). These are then used to calculate each joint angle. As the neutral position of the ankle is 90°, this means subtracting 90. Positive values for this angle therefore represent dorsiflexion, and negative values represent plantarflexion. The ankle joint angle is therefore calculated as follows:

$$\theta \text{ Ankle Joint} = 90^{o} - (\theta \text{ proximal segment} - \theta \text{ distal segment}).$$

Equation 2.3.

2.4.3. Subtalar joint motion: Eversion and inversion.

The subtalar or talocalcaneal joint lies within the vertical weight-bearing column between the heel and the tibia; between the concave under-surface of the talus, and the convex posterior surface of the calcaneum (Perry, 1992). The axis of the joint is obliquely orientated, inclined in the sagittal plane to approximately 45° (Root, Weed, Sgarlto, & Bluth, 1966). The joint runs in a primarily forward, but also upwards and medial direction, allowing the foot to tilt medially and laterally (Whittle, 2003). A large number of ligaments join the two bones together and to all adjacent bones; its ligamentous support consisting of the collateral ligaments, the interosseous ligaments and the talocalcaneal ligaments (Hall & Brody, 2005). This joint is in its neutral position when the posterior aspect of the heel lies vertical to the supporting surface of the foot. This neutral position becomes the baseline to determine foot and lower limb alignments (Norris, 2004). The functional importance of this joint lies in its capability to permit abduction and adduction of the hind foot; two movements essential to normal gait (Whittle, 2003). Furthermore, an important feature of the subtalar joint is its ability to perform tri-planar motion which is permitted by its oblique nature (Norris, 2004). Pronation of the foot is one such tri-planar motion, involving eversion of the calcaneus (frontal), abduction (transverse) and dorsiflexion (sagittal) (Norris, 2004). Pronation and supination are presented visually in Figure 2.8.

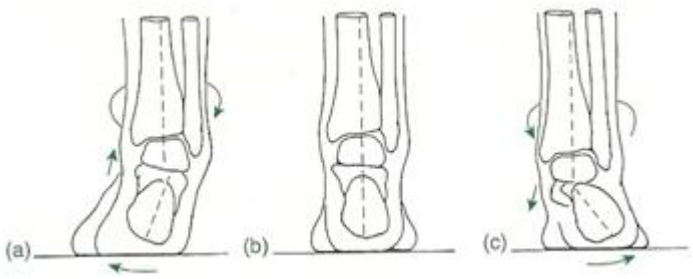


Figure 2.8. Position of the right subtalar joint in weight bearing, showing supination (a), neutral (b) and pronation (c). Source: Adapted from Norris (2004) p.182.

Inversion and eversion movements cause the foot to tilt on the obliquely orientated axis in a medial and lateral direction respectively (Perry, 1992). During one gait cycle, both movements occur throughout the swing and stance phases; however the motion during the latter is more significant due to its influence on the weight-bearing alignment of the proximal joints (Perry, 1992). The stance phase of running gait consists of three stages; heel strike, midstance, and toe-off. At the point of initial contact, the heel is slightly inverted, and the forefoot supinated (Whittle, 2003). This inversion is controlled by the actions of the tibialis anterior and posterior, with normal ranges between 0° to 25° (Kingston, 2000). During the first stage of heel-strike stance, following initial plantar flexion, the subtalar joint moves from an inverted to an everted position. This is a normal passive response to initial contact of the heel with the ground, and unlocks the midtarsal joint to produce a relatively flexible forefoot enhancing shock absorption and adaptation to irregularities in the ground floor surface (Nordin & Frankel, 2001). This is combined with the toes moving outward away from the mid line of the body (abduction) and the decreased distance between the toes and the lower leg (dorsiflexion). As mentioned, pronation is the term used to describe this combination of three movements. The movement of pronation causes the foot to roll medially, turning the sole of the foot outward and flattening the longitudinal arch. Due to the oblique axis of the subtalar joint, pronation and supination are therefore coupled with internal/external rotation of the lower leg (Norris, 2004). Normal ranges of subtalar joint eversion are known to be between 0° and 15°; above the latter value is defined as excessive (Kingston, 2000).

2.4.4. Motion of the foot: Abduction/adduction.

Movements of the ankle and foot joint are achieved through the combined actions of the talocrural joint, the subtalar joint, and the midtarsal joint (Perry,1992). The midtarsal joint consists of two articulations, the talonavicular joint and the calcaneocuboid joint. The midtarsal comprises joints which lie between each of the tarsal bones and their immediate neighbours. Together, the joints are combined to create a link between the hind foot and the forefoot, permitting a small amount of movement in all directions (Whittle, 2003). The movement of this joint is dependent on ligamentous and muscular tension to maintain its position and integrity (Hyde & Gengenbach, 2007). The ligamentous support is provided by

the actions of the calcaneonavicular, deltoid, calcaneocuboid and the dorsal talonavicular ligaments (Hyde et al., 2007). The peroneal muscles serve as both plantar flexors and evertors of the foot, and the peroneus longus abducts the forefoot in the transverse plane, thereby serving as a support for the medial longitudinal arch (Dutton, 2004). However, although stable, movement of the subtalar joint alters the alignment of the midtarsal joint of the foot (Norris, 2004). During subtalar joint pronation and supination, the two articulations of the midtarsal joint become hypermobile and hypomobile respectively (Hyde et al., 2007). Figure 2.9 illustrates the two movements of the foot: abduction and adduction.

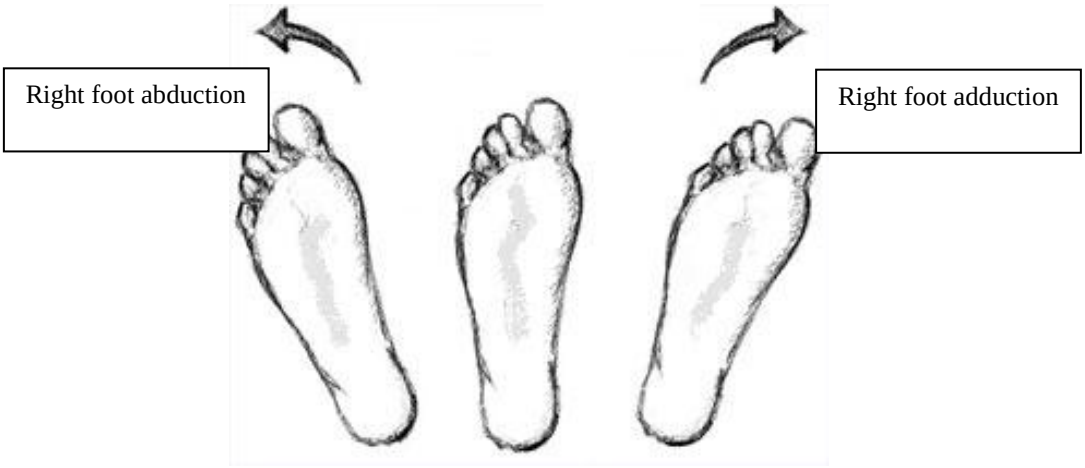


Figure 2.9. Abduction and adduction of the right foot. Source: Adapted from The Complete Foot Health Clinic, (2003).

2.4.5. Knee joint motion.

2.4.5.1. Sagittal plane movements: flexion/extension.

The knee is the largest and substantially most complex joint in the human body (Marieb & Hoehn, 2004). Although ostensibly a hinge joint permitting flexion and extension, the knee also allows frontal plane motion of abduction and adduction, and transverse plane internal and external rotation. Throughout gait, the knee undergoes movement in all three planes of motion, suggesting the knee is loaded in different areas at different points in time. As such, knee injuries and degradation can occur in various areas of the joint. This theory is considered in greater detail in Study 3. Knee movement in the sagittal plane involves flexion and extension, both of which occur at the tibiofemoral joint. The neutral position of the knee is the position where by no flexion or extension angle is calculated; from which up to 140° of flexion is possible in the uninjured human body (Norris, 2004). Knee flexion involves bringing the posterior surfaces of the upper and lower leg together, and is controlled by both the eccentric contraction of the quadriceps and the concentric action of the hamstring muscle groups (McGinnis, 2005). The sole knee extensor is the quadriceps femoris muscle of the anterior thigh; the most powerful muscle in the body (Marieb & Hoehn, 2007). During the stance phase of gait, the knee is extended at ground contact, with an average knee angle of less than 10 degrees, and then reaches peak flexion during mid stance (Whittle, 2003). This is often accompanied by initial plantar flexion of the ankle joint, with a net effect of acting as a shock absorber during the loading phase. The knee flexion during the mid stance phase is controlled by the eccentric contraction of the quadriceps, and reflects their role as shock absorbers (Richards, 2008; Novacheck, 1998). Finally the knee extends to a similar position of that achieved at ground contact, rapidly followed by ankle dorsiflexion to allow the toe to clear the ground.

2.4.5.2. Frontal plane motion: Knee abduction/adduction.

As described, the knee is predominantly a hinge joint, with motion permitted in both the frontal and transverse planes. Therefore, the majority of movement at the knee will take place in the sagittal plane of motion. However, although often debated, the joint does have to ability to permit a small degree of movement in the coronal plane (Richards, 2008). Although minimal, during loading there may be a slight degree of adduction or abduction of the joint, the direction of which is determined by the anatomical alignment of the knee; varus or valgus respectively (Richards, 2008). This is highlighted in the diagram below.

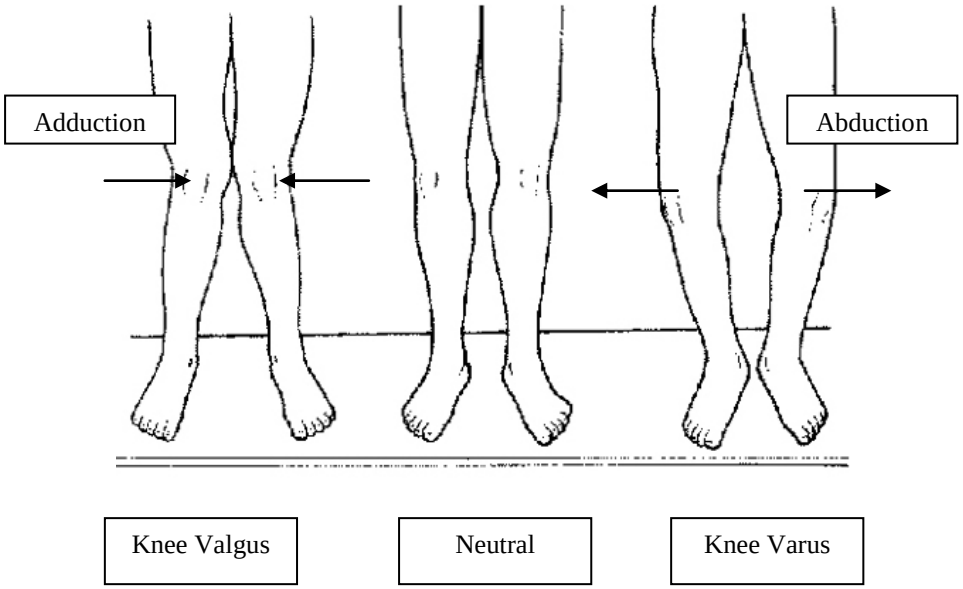


Figure 2.10 Illustration of the valgus and varus position of thee knee joints. Source: adapted from myorthosports.com.

However, as described, this deformation is controlled and therefore minimal. This is due to the control and stability of the joint, achieved through the action of ligaments (Whittle, 2003). The medial collateral ligament prevents adduction or valgus actions at the knee joint, and the lateral collateral ligament opposes abduction or varus actions. Both ligaments therefore act to stop their respective side of the knee from unfastening or opening up (Whittle, 2003).

2.4.5.3. Transverse plane motion: Knee internal rotation.

During gait, the knee joint permits a degree of movement in the transverse plane, predominantly dominated by the tibia rotating about the femur. As motion in this plane is less versatile than in the sagittal plane, important functional and pathological movements may be observed during these motions (Richards, 2008).

A common theme in the literature is that internal and external rotation of the knee is often coupled with flexion and extension of the joint during both stance and swing phases of gait. According to Hertling and Kessler, (2006) this is attributed to the structure and function of the joint and the femoral condyles. Figure 2.2 illustrates the condyles of the knee joint, and Figure 2.11 illustrates the association between sagittal plane motion and rotation of the knee joint. The medial condyle is curved and oblique in orientation; where as the lateral condyle is situated on the sagittal plane, and shorter in the anteroposterior direction than the medial. Due to the longer length of the medial articulating surface, as the tibia extends the lateral side of the joint completes its motion prior to the medial. The medial tibia then continues to move forward along the curved medial femoral condyle, whereas the lateral tibial joint surface undergoes a twist. The net effect is therefore a rotation of the tibia on the femur in an external direction. This process is then reversed throughout flexion of the tibia, as a combination of rolling and gliding movements of the femoral condyles occurs (Hertling & Kessler, 2006). This is supported by Norris (2004), who suggested that it is during the last 15° of extension during the stance phase of gait that the knee rotates. Additionally, knee internal/external rotation is often a consequence of the action of the cruciate ligaments; as the knee extends the cruciate ligaments tighten and are twisted in a direction to rotate the tibia (Hertling & Kessler, 2006).

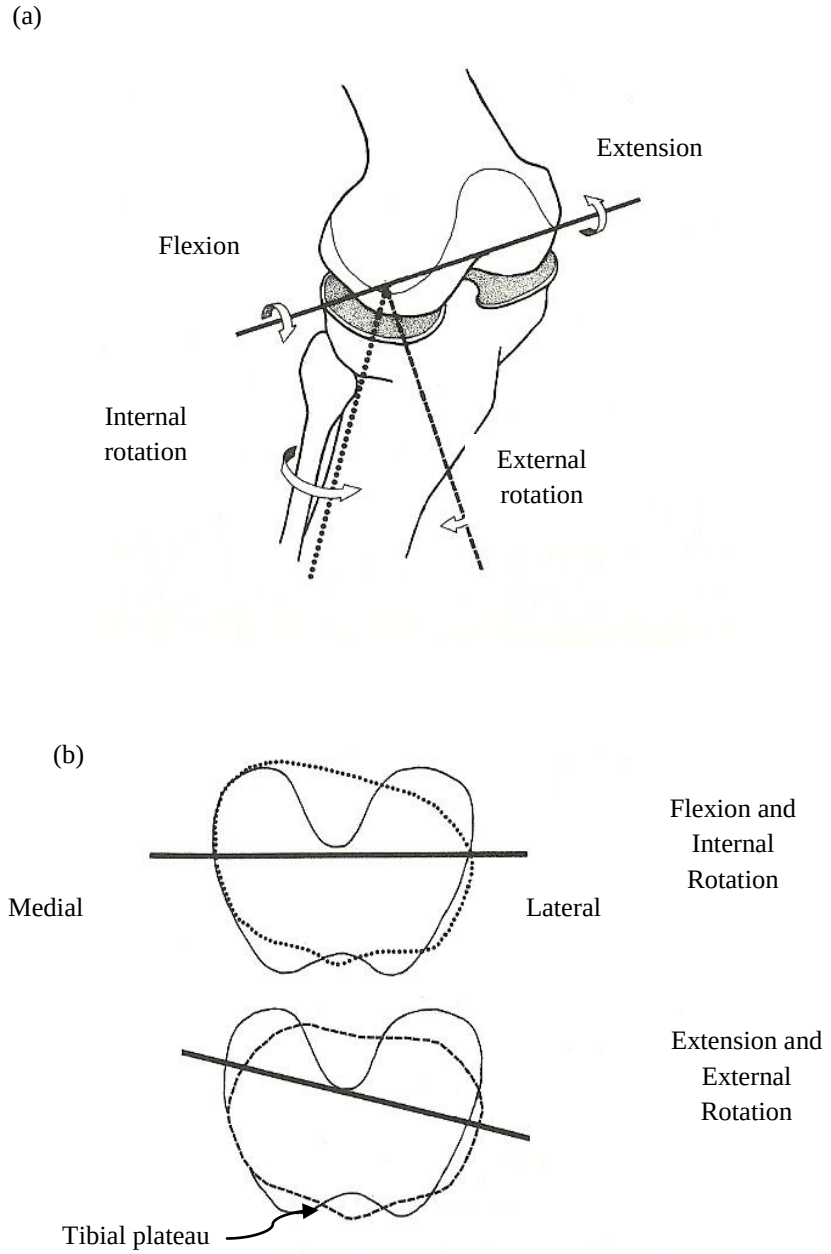


Figure 2.11a and b. Oblique view of the femur and tibia with the tibial plateau shaded (a) and superior view of the knee joint highlighting the position of the tibial plateau (dotted line) on the condyles during flexion and extension (b). Source: Adapted from Nordin & Frankel, (1989) p.121.

2.5. Kinetic Analysis.

"Kinetic variables are based on force – the cause of change in motion".

(McGinnis, 2005).

2.5.1. The ground reaction force.

In 1680, Borrelli was the first to measure the centre of mass of the body, and the first to consider ground reaction forces that act during gait (Richards, 2008). These are based on Newton's third law of motion. This is commonly stated as

"To every action, there is an equal and opposite reaction".

During static standing, a force is being applied to the person in equal and opposite magnitudes to body weight (Kirtely, 2006). This is known as the ground reaction force. In reality we are never completely static, and variations in body position will affect the exact location through which the ground reaction force acts (Richards, 2008). During both walking and heel strike running gait, the magnitude of ground reaction changes, showing peaks at both ground contact (peak impact force) and push off phase (peak active force) of the gait cycle (Kirtley, 2006). This force is distributed over the entire surface of the foot that is in contact with the ground; the centre of pressure thereby describing the theoretical movement of the applied ground reaction force throughout stance (Miller, 1990).

This ground reaction force is a vector quantity, whereby the resultant force can be reduced to components which describe the magnitude of the force acting along the three orthogonal axes (x, y, and z). This is highlighted in Figure 2.12. A typical convention is set up such that the x axis refers to the forces acting in the medio-lateral direction, the y axis indicates the anterio-posterior direction, and the vertical direction is illustrated by the z axis (Miller, 1990). A typical recording of all three components of ground reaction force during running is

highlighted in Figure 2.13. Among the vertical ground reaction vector, the graph illustrates the previously mentioned peaks associated with heel strike and ground contact and the push off phase. According to Miller (1990) the majority of research has focussed on this component during analysis of gait, due to the magnitude and distinguishable trace. Within this research thesis a specific focus has remained on the vertical component of the ground reaction force during gait. This is further supported in a later section, with specific focus on ground reaction force variables and injury (Section 3.1.3.1).

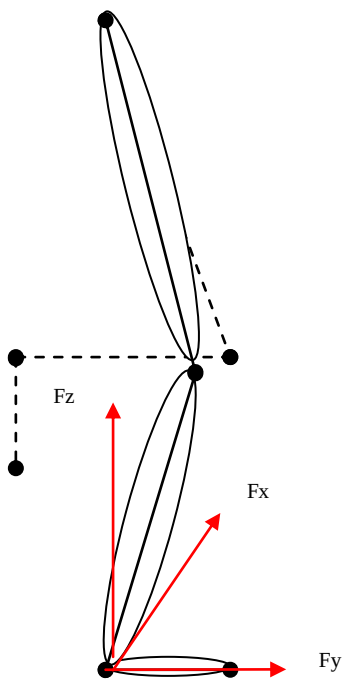
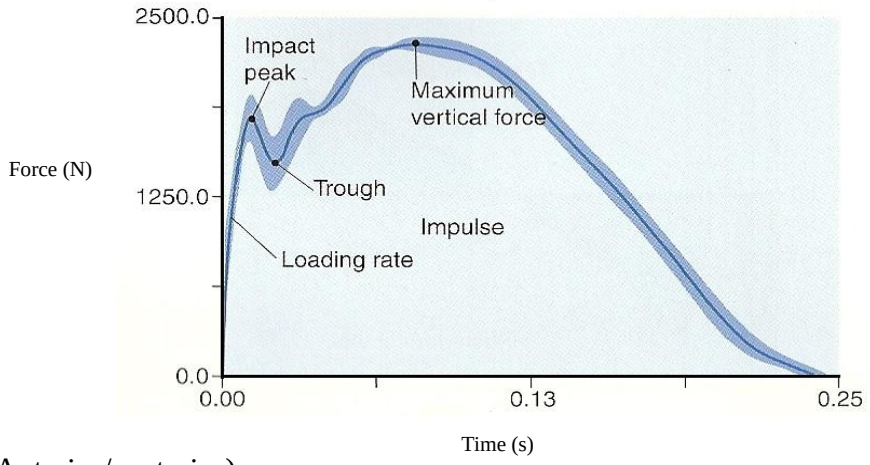
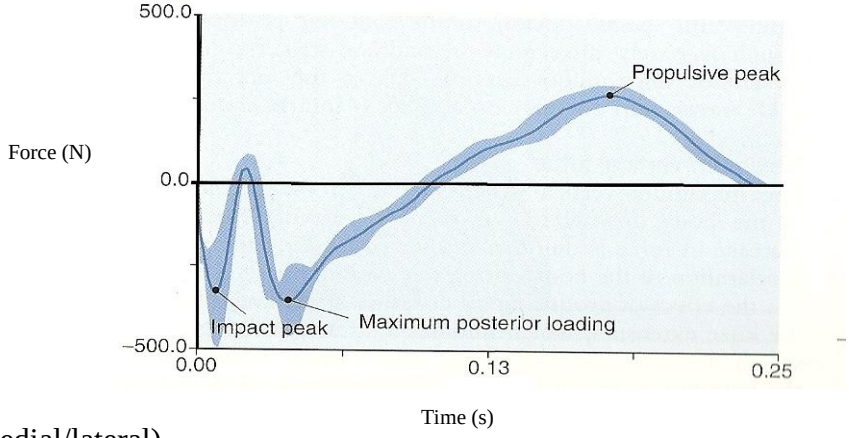


Figure 2.12. Diagram of the body during the stance phase of gait, illustrating the three components of ground reaction force (dotted line illustrates position of second leg).

(a: Fz, Vertical)



(b: Fx, Anterior/posterior)



(c: Fy, Medial/lateral)

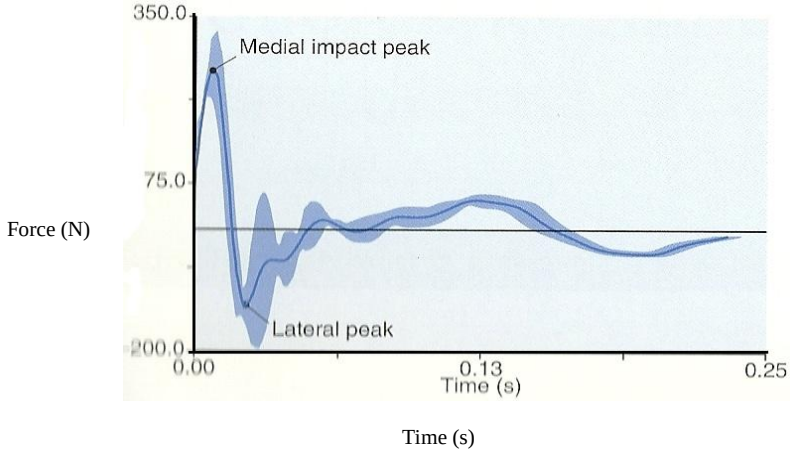


Figure 2.13a-c. Sample running ground reaction force traces for the rearfoot striker; illustrating the three components of ground reaction force. Source: adapted from Richards, 2008.

As well as the vertical component of the ground reaction force, Figure 2.13 illustrates both the anterior-posterior and the medial-lateral component. Following ground contact, the anterior-posterior component peaks in the opposite direction to the vertical ground reaction force, and is termed the braking phase (Munro, Miller, & Fuglevand, 1987). Following this, the second peak of the vertical component acts at a similar time to the positive peak of the anterior-posterior trace, termed the propulsion phase (Munro et al., 1987). The medial-lateral force trace illustrates minimal movement of the force in this direction. According to Miller (1990), the vertical component of the ground reaction force is a function of movement performance that is readily understandable, and therefore will take the focus of the ground reaction force assessment within this research thesis.

The loading rate of the vertical ground reaction force describes the gradient of the first peak, and illustrates the rate of the impact force development. This is calculated using the first central difference method as follows (Equation 2.4):

$$\textit{Loading rate} = \frac{(F_{final} - F_{initial})}{(t_{final} - t_{initial})}$$

Equation 2.4

Where F_{final} and $F_{initial}$ refer to the force points immediately after and prior to the point for which the loading rate is to be calculated, and t_{final} and t_{inital} are the time components for the respective force points.

2.5.2. Force platform accuracy and targeting.

Individual ground reaction forces during running can be determined using a force platform; a precision instrument using either strain gauges (Advanced Medical Technilogies Incorporated, Watertown) or piezo-electric quartz crystals (Kistler Instruments, Witerthur, Switzerland). The measuring surface is square or rectangular in shape, with the usual measurements of approximately 40 cm by 60 cm (McGinnis, 2005). Both methods convert ground reaction force into electrical signals (Kirtely, 2006). Overall, force platforms have

been suggested to provide an accurate method of ground reaction force assessment during movement. According to Gill and O'Connor (1997), the accuracy is however dependant on the method by which the force platform was mounted into the floor, as unstable force platforms can alter the signal produced.

The accuracy of force platform data was investigated by Gill and O'Connor (1997) who determined that vertical force measurement and reconstruction of the centre of pressure is acceptable with the use of an AMTI force platform, with low levels of interference between the different components of ground reaction force (Gill and O'Connor, 1997). However a limitation associated with kinetic force assessment using force platforms is the force plate targeting inaccuracy. According to Miller (1990), the major consideration when investigating ground reaction forces with a platform, is whether the data collected from a running trial is representative of the individuals normal gait. When undergoing an assessment of gait, it is common for individuals to alter the stride in attempt to contact the force platform with the desired foot, which may influence the quality of data produced.

According to one investigation by Grabiner and colleagues, force plate targeting did not significantly affect the ground reaction force or increase variability of the force profile among healthy males during walking trials (Grabiner, Feuerbach, Lundin & Davis, 1995). This result supported the earlier research by Patla and colleagues who showed no significant difference in ground reaction impulse between the conditions with and without visual cues for foot placement (Patla, Robinson, Samways, & Armstrong, 1989). In contrast however, Cuddeford and colleagues demonstrated a significant change in hip power when subjects were asked to target a force plate during walking compared to the trials performed with no visual cues (Cuddeford, Yack, Jensen, Peterson, Simonsen & Eichelberger, 1998). Similarly, in a study with experienced runners, Challis (2001) showed that trials including force plate targeting elicited significant differences in the peak magnitude of vertical ground reaction forces and a change in spatial coordinates over time (Challis, 2001). However, consistency was demonstrated in the ground reaction force traces, and the low standard deviations suggest that force plates are a reliable method for assessing ground reaction forces during gait, supposing no obvious stride alterations are made in attempt to contact the platform (Challis, 2001).

2.6. Joint Moments.

Shelburne, Torry & Pandy (2006) showed that the major contribution to medial knee joint loading was the orientation of the ground reaction force vector due to its component in joint moment calculation. The following section describes the calculation of moments at the joints of lower limb, and the relevant application of the ground reaction force components.

Moments that act about the joints during running are calculated as the product of the joint force and the distance of that force from the joint centre location (Hay, 1993). As such, details of the internal forces acting about the joints are required.

2.6.1. Anthropometry.

The method of inverse dynamics combines the study of kinetics and kinematics to obtain quantitative values for joint moments during the support (stance phase) of gait (Richards, 2008). This approach involves a series of Newton's equations and the construction of a link-segment model of the lower limb, disarticulating each segment to form rigid bodies (Winter, 1990). Basic rigid body analysis of the lower limb involves the individual segments of the foot, shank, thigh and trunk, each with a uniform density and the mass located at a single point along the length (centre of mass). Identification of centre of mass (CoM) location and magnitude can often be quantified using precise measurements from magnetic resonance imaging (Richards, 2008). However due to lack of availability and cost, the majority of research relies on measurements performed using cadavers (Demster, 1955). The first stage of this process involved dissection of the cadaver (frozen limb), to disarticulate the individual segments. Simply, the single segments were then weighed to determine the mass, and balanced on a single point to determine the location of that segment centre of mass (Dempster, 1955). The latter calculation was then expressed as a percentage of length from the distal end (Kirtley, 2006). According to Richards (2008), the measurements suggested by Dempster in 1955, although categorised estimates, are still considered the ideal anthropometric values for body segment parameters.

It is however important to consider that the estimation of joint centre of mass through anthropometric measurements has a few limitations. Although applied to calculations among a range of human participants, values based from the work of Dempster and colleagues do not account for vast cases of atypical distribution of body mass (Cotton, Vanoncini, Fraisse, Ramdeni, Demircan, Murray and Keller, 2011). In general, the calculations of centre of mass location account for variations within a relatively small category of participants including healthy adult Caucasians. As such, the higher the deviation from nominal body mass distribution, the higher the modelling error (Cotton et al., 2011).

2.6.2. Joint forces.

2.6.2.1. In-vivo measurements.

Within each joint in the human body, a state of equilibrium exists where the external forces are balanced by the internal joint forces (Richards, 2008). Understanding these forces provides information regarding muscle mechanics, physiology, motor control, musculoskeletal mechanics and neuro-mechanics (Kjaer, 2003). As such, within the field of biomechanics, it is of interest to quantify the patterns of force acting on the muscles, ligaments and tendons in the joints (Roberston, 2004). Developments in technology and surgical implantation procedures made it possible to measure forces internally, which lead to an application of in-vivo techniques for studying the loading of tendons during human locomotion (Kjaer, 2003). Early research therefore investigated methods to successfully quantify these internal forces including internal force transducers; however an abundance of limitations and negative reception of their use made them subject to concern.

Within an Achilles tendon study, although a quantification of the force was obtained, the output only provided information on the activity of the Achilles tendon, with no data from the forces of the triceps surae muscle groups (Kolt & Snyder-Mackler, 2003). Furthermore, difficulties associated with calibration of the transducers exist, as calibration will always be specific for the single tendon investigated (Meyer, Nyffler, & Geber, 2004). This requires calibration to be intraoperative at implantation, or precise measurement of the tendon

diameter, both of which require the use of specialised equipment and expertise, alongside the potential discomfort to the participant (Meyer et al., 2004). Finally, the placement of equipment internally will potentially disturb the observed motion of the limb (Robertson, 2004). Therefore, although such methods have the potential to provide accurate information regarding internal forces, measurement of forces in-vivo requires invasive methods that are subject to limitations. As such, these techniques are unlikely to be widely accepted as a means of quantifying internal joint forces during motion (Robertson, 2004).

2.6.2.2. External measurement of joint forces.

During the stance phase of gait, several forces act on the joints of the lower limb; combining to provide an overall vertical and horizontal joint force. A diagram of the forces acting at the ankle joint is displayed in Figure 2.14.

As previously described, every single body segment has a mass, which in turn has acceleration due to gravity. As such, the weight (mg) of that segment can be assumed as a combination of its mass (m) and the acceleration due to gravity ($g=9.81\ m.s^{-1}$) (Kirtley, 2006). This weight of each segment is then assumed to act vertically down at its centre of mass; the location of which is determined through cadaver data (Dempster, 1955). Therefore, as the weight of the foot acts at the centre of mass, so the ground reaction forces act at the segments centre of pressure. As such, when the foot is in a plantarflexed position during stance, the weight and the ground reaction forces act at different places (see Figure 2.14) (Kirtley, 2006). Finally, as the ankle-foot segment has been disarticulated from the body, there is an additional force acting, known as the joint reaction force (Kirtely, 2006). Using Newton's laws of motion (specifically, force = mass x acceleration), these forces are considered to be acting along a reference frame, and can be resolved in both the vertical and horizontal directions.

Forces acting on the ankle and foot:

Horizontal: $R_{x\ ank} + F_x = ma_x \longrightarrow R_{x\ ank} = ma_x - F_x$

Vertical: $R_{y\ ank} + F_y - mg_{foot} = ma_x \longrightarrow R_{y\ ank} = ma_x - F_y + mg_{foot}$

Equation 2.5

Where $R_{x\ and\ y}$ and $F_{x\ and\ y}$ are the ankle reaction forces and the components of GRF in both the horizontal (x) and vertical (y) components respectively.

As displayed in Figure 2.14, when there is no horizontal acceleration acting on the foot segment, this joint is in a static position. As such $a_{x\ and\ y}$ will be close to zero, negating the ma component. The horizontal ankle force R_x is therefore equal and opposite to F_x. These methods are also applicable to the proximal joints of the lower limb. The equations below illustrate the overall forces acting at the knee and hip joint during a static position of the limb.

Forces acting on the shank and knee:

Horizontal: $R_{x\ knee} + R_{x\ ank} = ma_x \longrightarrow R_{x\ knee} = ma_x - R_{x\ ank}$

Vertical: $R_{y\ knee} + R_{y\ ank} - mg_{shank} = ma_x \longrightarrow R_{y\ knee} = ma_x - R_{y\ ank} + mg_{shank}$

Equation 2.6

Forces acting on the thigh and hip:

Horizontal: $R_{x\ hip} + R_{x\ knee} = ma_x$ $\longrightarrow$ $R_{x\ hip} = ma_x - R_{x\ knee}$

Vertical: $R_{y\ hip} + R_{y\ knee} - mg_{thigh} = ma_x$ $\longrightarrow$ $R_{y\ hip} = ma_x - R_{y\ knee} + mg_{thigh}$

Equation 2.7.

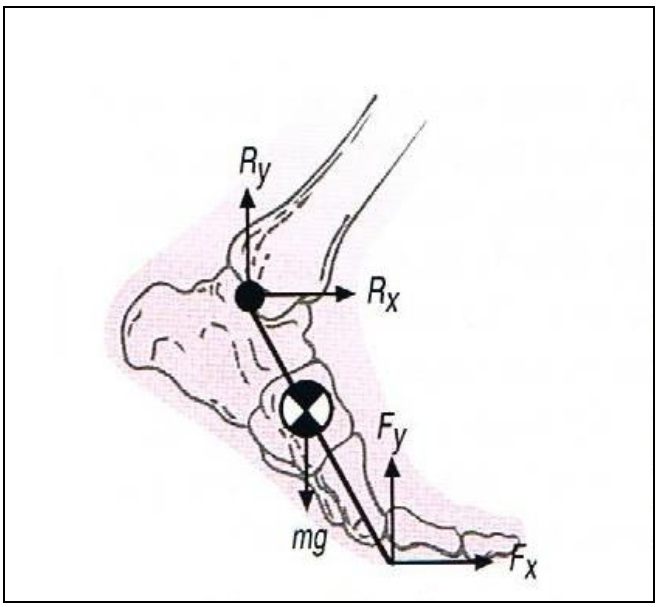


Figure 2.14. Free body diagram of the foot and ankle complex, illustrating the forces acting at the joint during a static position. Source: Kirtely (2006) pp.122.

2.6.3. Joint moments.

The method of inverse dynamics simply implements a rigid-linked segment model, used to calculate the net muscle moment acting at the ankle, knee and hip joints (Farley et al., 1998). As such, forces applied to the joint through the actions of the muscles need to be considered. Using the example of the ankle-foot segment, concentric actions of the gastrocnemius-soleus and the anterior tibialis muscle groups cause the foot to plantar-flex and dorsi-flex respectively. This in turn produces a moment at the joint acting negatively in a clock-wise or positively in an anticlockwise direction (Richards, 2008). In simple terms, a moment acting at a joint is calculated by multiplying the magnitude of the total force acting through the joint, by the perpendicular distance to the fulcrum (CoM), known as the moment arm (Whittle, 2003). Therefore within the calculations, the magnitude of each force is multiplied by the perpendicular distance from the location of the force to the centre of mass. As such, the moment arm of the segment weight is zero as it passes directly through the segment COM. The moment acting at the joint is calculated using the following equation:

$$M = I\alpha,$$

Equation 2.8

where I is the moment of inertia, and α is the angular acceleration of the segment (Kirtley, 2006). Therefore, assuming that the joint is frictionless, and that air resistance is negligible, the formula for calculating the ankle joint moment is as follows:

Forces and moment at the ankle joint:

Horizontal: $R_{x\ ank} = ma_x - F_x$

Vertical: $R_{y\ ank} = ma_x – F_y + mg_{foot}$

Ankle Moment: $M_a = F_y d_x + F_x d_y – Ryp_x – R_x p_y + I\alpha$

Equation 2.9.

Where d_x d_y p_x p_y are the distances from the respective forces to the joint centre of mass.

Similarly, following calculation of the joint reaction forces and the moment acting at the ankle joint, moments can then be determined at the proximal joints of the lower limb; the knee and the hip.

Forces and moment at the knee and hip joint:

Horizontal: $R_{x\ knee} = ma_x - R_{x\ ank}$

Vertical: $R_{y\ knee} = ma_x – R_{y\ ank} + mg_{shank}$

Knee Moment: $M_k = R_{y\ ank}p_x + R_{x\ ank}p_y - R_{y\ knee}q_x - R_{x\ knee}q_y – I\alpha$

Equation 2.10.

Horizontal: $R_{x\ hip} = ma_x - R_{x\ knee}$

Vertical: $R_{y\ hip} = ma_x – R_{y\ knee} + mg_{thigh}$

Hip Moment: $M_h = R_{y\ knee}q_x + R_{x\ knee}q_y - R_{y\ hip}t_x - R_{x\ hip}t_y – I\alpha$

Equation 2.11.

As described, joint moments simply symbolise where the resultant force is acting at the joint during motion. This information is important in clinical assessments and research projects, as it highlights the presence of joint malalignment and uneven force distribution at the joint (Whittle, 2003).

2.7. Knee Joint Stiffness.

In previous sections, the relative influence of kinematic variables, kinetic variables, and joint moments have been considered in their association with the development of lower limb injuries and over use conditions. Knee joint stiffness is an additional biomechanical variable that has been similarly associated with running related conditions, due to a relative reduction in joint range of motion.

2.7.1. Introduction to stiffness.

Hopping, trotting and running are a variety of different gaits used by legged animals to move from one place to another. Although all animals vary in body shape and dimensions, each gait permits movement across the ground similar to the motion of a bouncing ball, as the musculoskeletal system behaves like a spring (Farley, Houdijk, Strien, & Louie, 1998). This occurs as the muscle-tendon complex of the lower extremity alternately stretch and shorten using the elastic potential of the muscles and tendons (Kuitunen, Komi & Kyrolainen, 2002). According to Gunther & Blickhan (2002), the spring-mass model provides a well accepted theoretical basis on which to construct the global dynamics of fast legged locomotion under gravity. These varied gaits of animals can consequently be modelled using a spring-mass system, and as a result, the concept of stiffness has its origins in physics. Hooke's Law applies to all deformable bodies which store and return elastic energy, providing their shape is not permanently changed. Hooke's law is defined as follows:

$$\sigma = k\gamma,$$

Equation 2.12.

where the force required to deform a material (σ), is proportional to the spring constant (k), and the distance the material is deformed (γ) (Menard, 2008). The leg of an animal is often modelled as a spring supporting the mass of the body (Figure 2.15). The constant (k)

describes the stiffness of the spring and mass system (Butler, Crowell, McClay Davis, 2003), and as a result represents the average overall stiffness of the musculoskeletal system during the ground contact phase of gait (Farley et al., 1998).

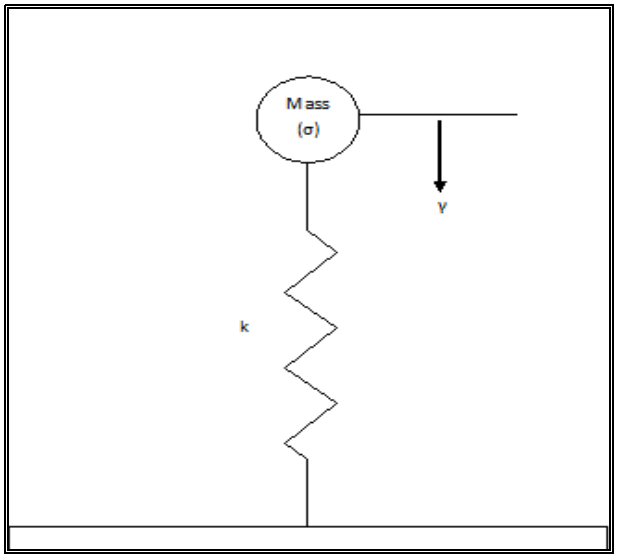


Figure 2.15. A spring-mass model representative of a single linear leg spring, illustrating "mass" equivalent to body mass or force (σ), deformation of material (γ), and leg spring (k). Source: Adapted from Butler et al., (2003).

In its simplest terms, stiffness is the relationship between the deformation of a body and a given force; the force acting on the leg divided by the change in length of the leg (Butler et al., 2003; Kuitunen, Komi & Kyrolainen, 2002). It is the ability of the system to resist the applied stretch or force (Kuitunen et al., 2002). As displayed in Figure 2.15, the leg is often modelled as a spring supporting the mass of the body during gait. When studying the mechanical behaviour of the musculoskeletal system, the advantage of the spring-mass system to identify leg stiffness is its simplicity, as it uses only one idealised spring to model

the behaviour of the body. However, it has been suggested that this model ignores the mechanisms of this multi-spring system (Kuitunen et al., 2002). According to Farley et al (1998), components other than mass are known to affect the behaviour of the leg spring, and the changes in spring length correspond to the alternating flexion and extension motion of the joints. In a multi-jointed system such as the human skeleton, overall stiffness is dependent on the specific torsional stiffness of each joint as well as the geometry of the musculoskeletal system (Farley et al., 1998). This notion was later supported by Gunher and Blickhan (2002) who suggested that the elasticity (and therefore stiffness) of the leg during running and jumping movements can be identified on the joint level, and global leg stiffness may emerge from local elasticity established by the appropriate joint torques. As a result, overall stiffness of the lower limb can be determined by the individual magnitude of the torsional stiffness of the ankle, knee and hip joints.

2.7.2. Knee contribution to overall leg stiffness.

As described above, all joints of the lower limb contribute to overall leg stiffness. However, it has previously been suggested that the stiffness of the knee joint is the greatest contributor to leg stiffness. While linear stiffness is defined as linear force per linear displacement, knee joint stiffness is operationally defined as the ratio of the change in torque to a change in knee angle (Oatis, Wolff & Lennon, 2006). In an assessment of lower limb stiffness, Laughton, McClay and Hamill (2003) illustrated forefoot strikers to present greater knee joint stiffness than rearfoot strikers. Similarly, overall leg stiffness was greater in the forefoot group, whereas ankle stiffness was greater in the rearfoot group, suggesting knee joint stiffness has a greater contribution to overall leg stiffness than ankle joint stiffness (Laughton et al., 2003). This notion was further supported in a study investigating the effect of knee stiffness on leg stiffness during hopping. Hobara, Muraoka, Omuro, Gomi, Sakamoto, Inoue and Kanosue (2009) performed a multiple regression analysis which revealed knee stiffness to explain more of the variance in leg stiffness than did the torsional stiffness of the ankle or hip. According to the results of the discussed study, the spring-like behaviour of the leg during hopping is adjusted by changing the knee joint stiffness, while the stiffness at the ankle joint remains constant (Hobara et al., 2009). Finally Arampatzis, Bruggemann and Metzler (1999)

illustrated that with increasing velocity, as well as increases in overall leg stiffness, larger changes were observed in the spring stiffness of the knee joint compared with that at the ankle joint, indicating that the increase in leg stiffness is caused in the most part by the increase in knee joint spring stiffness.

During movement, a degree of stiffness is required to achieve optimal utilization of the strength-shortening cycle, resulting in efficient use of the stored elastic energy in the musculoskeletal system during loading (Butler et al., 2003). Therefore, a possible reason for the lack of change in ankle stiffness compared to that at the knee, is the ankle joint's inability to tolerate the applied load and utilise the stored elastic energy as efficiently as the knee joint (Hobara et al., 2009). Therefore, when compared to the ankle, the knee joint appears to be a greater contributor to overall leg stiffness.

Within this thesis, a specific focus has been on running related injuries and debilitating conditions that affect the knee joint. As illustrated above, the knee joint is the major contributor to overall leg stiffness; and as such knee joint stiffness will be investigated in its potential association with overuse conditions.

2.8. Muscle Strength.

2.8.1. Skeletal muscle structure and function.

Skeletal muscle is the tissue accountable for generating the forces that produce movement of joints during human motion. This tissue accounts for approximately 40 to 45% of total body mass (Hargreaves and Hawley, 2003). Within the human body, each skeletal muscle is a separate organ, composed of thousands of multinucleated elongated cells named muscle fibers (Tortora & Grabowski, 2003). It is the arrangement of structure and subsequent actions of elements within each of these muscle fibers that are responsible for each muscular contraction. The microstructure of a skeletal muscle fiber is displayed in Figure 2.16.

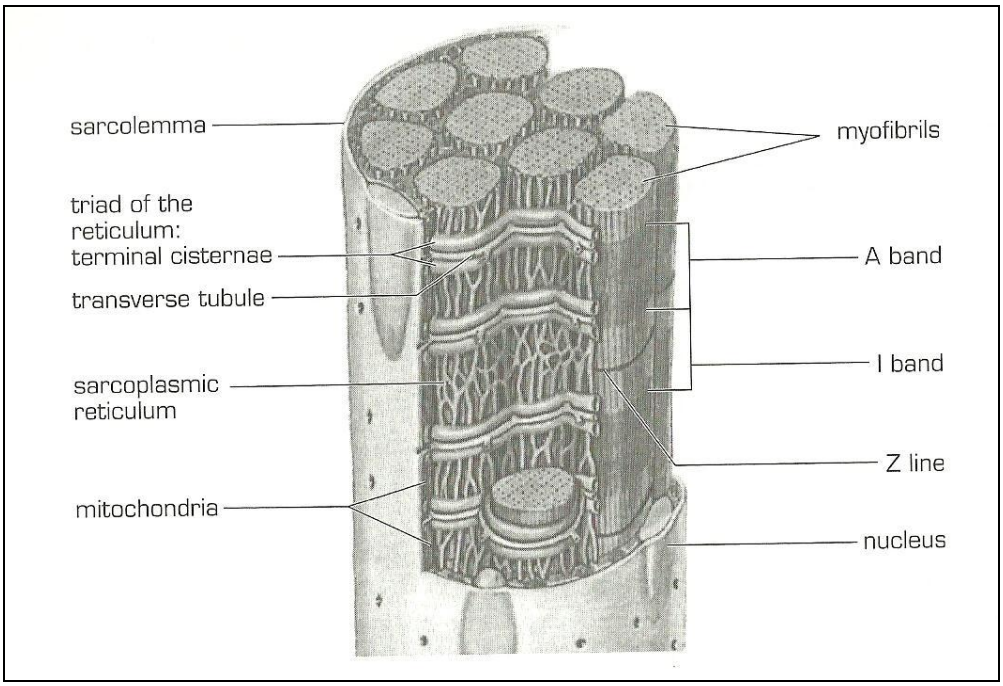


Figure 2.16. Image displaying microstructure of a single skeletal muscle fiber. Source: Hargreaves & Hawley, (2003).

Each muscle fiber often extends the length of the skeletal muscle; measuring from 10 to 100 micrometers in diameter and up to 750,000 micrometers in length. The predominant feature of each fiber is the numerous myofibrils; cylindrical shapes within the membranous network of the fiber called the sacroplasmic reticulum (Sherwood, 2007). These myofibrils constitute 80% of the entire muscle fiber, and contain the contractile apparatus; the thick (myosin) and thin (actin) protein filaments organised into contractile units (Hargreaves and Hawley, 2003). It is a cross bridge interaction between these two protein filaments by means of the sliding filament mechanism that causes muscle contraction (Sherwood, 2007). Each muscle is stimulated by motor neurones from the somatic nervous system which instigates an action potential propagated along the fiber. This in turn causes a release of calcium from the sarcoplasmic reticulum which allows the myosin head to attach to the actin filament (cross-bridge formation). The thin actin filaments then slide along the thick myosin filaments, undergoing a process known as the sliding filament mechanism. This produces an overall shortening of the sarcomere, the whole muscle fiber, and the entire skeletal muscle; initiating a single muscular contraction (Tortora & Grabowski, 2003).

2.8.2. Quadriceps Femoris Muscle.

During running, the movements of the lower limbs are predominantly controlled by actions of the quadriceps and hamstring muscles. Concentric contraction of the quadriceps causes a shortening of the quadriceps femoris muscle, producing an extension of the lower leg at the knee joint. Conversely, concentric action of the hamstring muscle group causes shortening of this muscle, producing a flexion movement at the knee joint (Robergs & Keteyian, 2003). Both movements occur countless times in each body during the course of one day, as each movement is inherent within the gait of both walking and running.

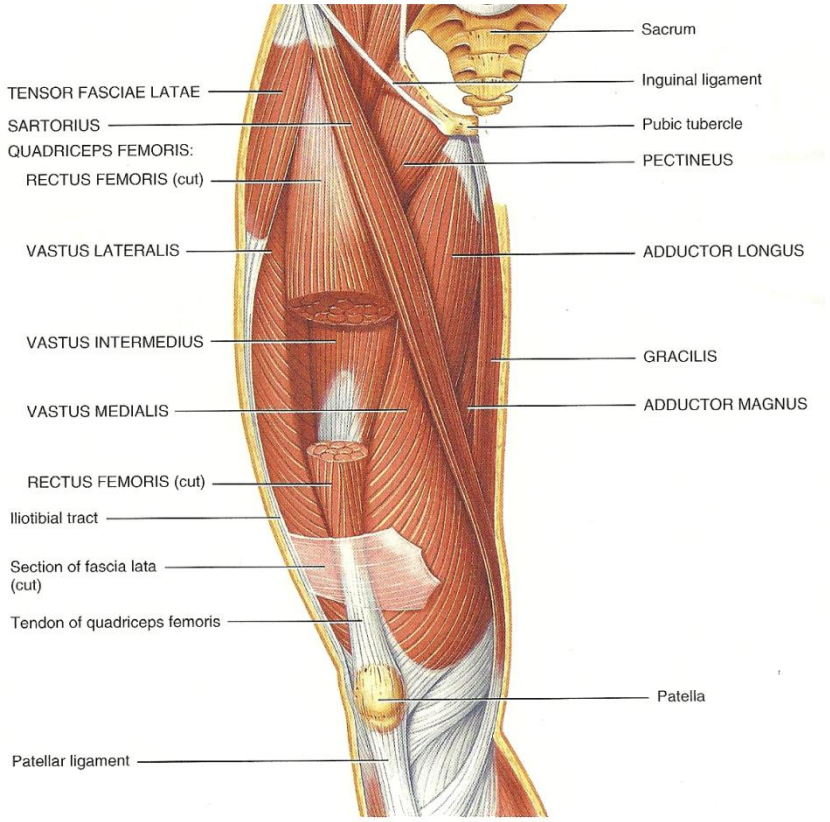


Figure 2.17. Image of the muscles acting around the knee joint, including the main contributors of the quadriceps femoris muscle (Source: Totora & Grabowski, 2003).

Figure 2.17 illustrates the main muscles acting around the knee joint. The quadriceps femoris is the largest muscle mass covering anterior, medial and lateral aspects of the femur, and functions as the greatest extensor of the leg. This composite muscle group is made up of the vastus medialis, the vastus lateralis and the vastus intermedius muscles. These are recognised as major contributors to patellofemoral joint function, and act to facilitate restraint of the patella in medial and lateral directions respectively. The rectus femoris muscle is the rounded anterior head of the quadriceps, and appears as a separate muscle despite its insertion (Jenkins, 1998). The insertion of all four muscles of the quadriceps femoris is at the patella tendon which inserts to the patella and through the patella ligament to the tibial tuberosity (Tortora & Grabowski, 2003). The rectus femoris muscle arises from the anterior inferior

iliac spine, and crosses both the hip and the knee joint. This muscle has varied functions; extends the lower leg and assists to flex the thigh on the pelvis. The vastus muscles serve to extend the leg at the knee joint, and facilitate restraint of the patella in the medial and lateral directions (Totora & Grabowski, 2009). Subsequently, weakness of these vastus muscles manifests as an abnormal position or motion of the patella with chronic instability of the knee joint.

2.8.3. Muscle strength.

Muscle strength is an essential component of most aspects of movement and performance, and is underpinned by muscular, neural and mechanical factors. Within the clinical setting the use of the term strength is almost always representative of muscle torque (Clarkson, 1999). Torque is the propensity of a force (muscle tension) to turn a lever (limb) about an axis of rotation (joint). It is measured as the function of the muscle force proportional to cross sectional area and the biomechanical advantage of the moment arm or lever system. Therefore measurement of muscle torque during isolated joint movements provides a value for the strength of the muscle under consideration.

The bracket of muscular strength is commonly divided into endurance and strength. Strength is the human ability to exert a physical force, and is measured as the maximal torque one can exert in a single maximum voluntary contraction (MVC), when the type of muscular contraction, limb velocity and joint angle are specified (Kumar, 2004). Conversely, muscular endurance is the ability to maintain a force over a period of time or a set number of repetitions or contractions (Rybski, 2004).

There are two basic categories of muscular contraction, dynamic (isotonic and isokinetic) and static (isometric); each is classified according to the mechanical changes that occur in the muscle during action. Isometric contractions occurs as tension is developed but the length of the muscle remains constant (Robergs & Keteyian, 2003). Both isotonic and isokinetic involve a change in the length of the muscle, either shortening (concentric) or lengthening (eccentric), however an isokinetic contraction produces movement at a constant speed.

Muscle actions generate tension that is transferred to bone and enables movement. Muscles are categorised according to their major role in producing this movement. An agonist muscle or prime mover is a muscle or muscle group that makes the major contribution to movement at the joint, for example the quadriceps muscle. Alternatively, an antagonist muscle relaxes as the prime mover acts, working in opposition to the agonist muscle (Clarkson, 1999). Examples of an antagonist and agonist working in opposition are the biceps and triceps muscle through flexion and extension of the elbow, or the quadriceps and hamstring muscles through flexing and extending at the knee joint. The strength of each muscle is different, and underpinning the choice of muscle action to examine must be the activity specific component under investigation (Armstrong, 2007).

2.9. Footwear.

2.9.1. Shoe design and function.

For the last three decades, the footwear biomechanics community have been asking fundamental questions regarding the association between footwear, performance and the development of injuries and debilitating conditions. To a large extent, this has been market-led due to the consistent increase in the number of people participating in jogging for both recreational and competitive purposes (Norris, 2004).

While footwear aims to protect and support the foot, the extensive range of shock-absorbing properties and motion control techniques cause running shoes to surpass what one may expect of an ordinary shoe. In 1980, the shoes gained the top rating in Runner's World Magazine were characterised by price, weight, shock absorbency and durability (Shorten, 2000). However since this time, great changes have been seen in the design of running shoes, and contemporary designers focus on the anatomy and movement of the foot-ankle complex, with shoes readily available in all styles and colours (Shariatmadari et al., 2010). It has now a common phenomenon that most running shoe manufacturers will include a method of anti-pronation or stability within the running shoe selection. Each running shoe may now have as many as 20 parts to it, as is illustrated in Figure 2.18 (below). Although the design of running shoes has become increasingly intricate, controversy still exists regarding the ability to create a shoe that will be suitable to the unique anatomy of each individual.

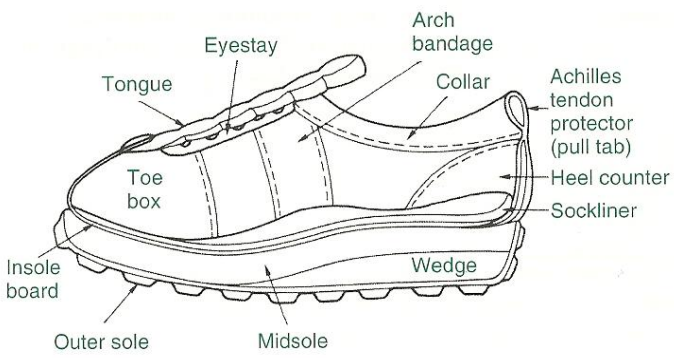


Figure 2.18. Components of the running shoe. Source: Adapted from Norris, (2004 p. 292).

The majority of the following theory is based on that of Norris, (2004), who described the components of running shoes in great depth (Figure 2.18). The purpose of the outer sole is to provide traction, tread, flexibility and grip. Above this is the insole board, midsole and the wedge, generally made from ethyl vinyl acetate (EVA) or polyurethane, which provides cushioning and foot control during the stance phase of running. The heel counter is often made from a hard thermoplastic material, which exists to counter rearfoot motion or excessive pronation of the subtalar joint. It is commonly the wedge and the heel counter that are adapted to form the motion control aspect of stability shoes (Norris, 2004). The shoe upper section is designed to contour the foot, and will vary depending on the anatomical width and depth of the foot, generally controlled by the lacing system.

2.10. Summary of literature review and general thesis rationale.

Running is a popular recreational activity among the mature female members of the population. However the risk for the development of running related injuries and conditions is high, with the risk for certain conditions becoming greater with age. These injuries and conditions seemingly dominate at the knee joint. Due to the association with gender and age, the specific condition of knee joint osteoarthritis is among the most frequently seen among this mature female group.

Although demonstrated as a high risk group, there is a lack of research regarding injuries and evidence of causative factors among this cohort. It is speculated that advancing age causes anatomical and physiological changes to occur, that may heighten an individual's susceptibility to the development of injuries and conditions, although there is a lack of research based on the specific group of mature females to confirm this. It is further considered that post menopausal females may be at risk of developing degenerative conditions at the joints, due to the reduction in oestrogen and a change in the catabolic and anabolic activity within the bones

It is speculated that as the risk of injuries and debilitating conditions appears to increase with age, certain biomechanical changes are expected that could place this specific group of mature females at risk. Due to the current lack of definitive evidence regarding running styles among mature females, investigation into changes in gait is justified.

Peak rearfoot eversion and knee internal rotation have consistently been linked with running injuries, although this has not been extensively documented among the more mature members of the population. Kinetically, loading rates and impact forces have been associated with overloading during running, and knowledge of joint moments provides information regarding the influence of these forces on the joints. Knee joint stiffness has been associated with osteoarthritis; however the cause – consequence relationship appears equivocal. It has been suggested that reductions in muscle strength may occur with age, and therefore could provide insight into possible injury mechanisms among this group. With a lack of previous research regarding running styles among mature female runners, the investigation of biomechanical changes with age is warranted.

It has been postulated that a possible management strategy for injuries and running related conditions, is through the prescription of footwear. Although not without contention, cushioning properties in shoes have been shown to reduce impact forces and loading rates. Similarly motion control devices have been found to limit rearfoot motion and internal rotation of the tibia and knee among runners. However, the potential of these footwear interventions to influence biomechanical risk factors among the mature members of the running population have not been well documented.

2.10.1. Long term study objective.

The main objective of this research project is to determine whether age related biomechanical changes occur in running gait among the female population of runners. Following this, the effectiveness of management strategies in the form of footwear control will be investigated specific for mature female runners. Finally, an investigation into the current condition of the knee joint of the mature females will occur, to further substantiate biomechanical variables as risk factors for the development of osteoarthritis.

2.10.2. Scope and boundaries of the study.

This research project describes and evaluates the biomechanical changes to running gait that occur with increasing age among active females. Furthermore, a second study will consider possible physiological factors will be considered in their contributory role to these biomechanical changes. This project then investigates the role of footwear in injury and osteoarthritis prevention and management, and considers biomechanical variables of gait as diagnostic tools. In the final study, the use of magnetic resonance imaging allows an investigation of diverse techniques to identify degenerative changes that occur at the knee joint with increasing age among females.

2.10.3. Study Overviews.

2.10.3.1. Study One.

Study one is an investigative study, which aims to directly compare the running gait of young and mature females to investigate whether mature females demonstrate gait characteristics that may predispose them to the development of running related knee injuries and overuse conditions. All subjects were assessed for the presence of knee joint injuries or osteoarthritis through the use of a validated knee osteoarthritis observation survey (KOOS). Both kinetic and kinematic analysis occurred to investigate variables associated with injuries including rearfoot eversion, knee internal rotation and loading rate. Knee joint moments were identified in the frontal plane to assess the risk of osteoarthritis. Subjects performed this assessment in their own running footwear.

2.10.3.2. Study Two.

Study Two involved further investigation of the biomechanical differences identified in Study One, with an additional focus on the influence of motion control footwear. Similar to Study One, a direct comparison between mature and young female runners took place. In addition, a current commercially available motion control shoe from a well known retailer was included as a test condition to assess the ability of a motion control shoe to reduce specific observed differences in running biomechanics between mature and younger female runners. Additional assessments of knee joint stiffness and muscle strength were included in the investigation, to develop the understanding of age related differences.

2.10.3.3. Study Three.

Study Three involved an intervention study. Based on the results of Study One and Study Two, the aim was to assess the effect of both a medial and lateral wedge, and a combined orthotic intervention with both medial and lateral support, on the running gait of mature

females. The changes in footwear conditions were based on each variable identified in the previous studies and an intervention multi-purpose orthotic footwear condition incorporating both a medial and lateral wedge was included to aim to control rearfoot motion and the knee external adductor moment among mature female runners.

2.10.3.4 Study Four.

Study four occurred at a similar time to Study Three. This supplementary investigation aimed to support the original theory that mature female runners are at a higher risk for the development of debilitating conditions, specifically knee joint osteoarthritis. A magnetic resonance imaging (MRI) scanner was used to scan the knees of each mature female, to identify any early signs of osteoarthritis present in the joint. These results were then correlated with previous surveys of osteoarthritis presence. Further correlation tests were performed to investigate the previously proposed association between specific biomechanical variables of gait and the development of injuries and debilitating conditions at the knee joint.

Chapter 3.

"A Biomechanical Comparison of the Running Gait of Mature and Young Females"

3.1. Introduction.

3.1.1. Running and the risk of injury.

The number of females participating in running is increasing, augmented by the overwhelming concern for the growing epidemic of obesity (Rose, Birch, & Kuisma, 2011). As a consistently available form of physical activity, middle and long distance running is a relatively simple means of increasing overall fitness (Birrer & O'Connor, 2004) and these participation rates have been attributed to the benefits associated with physical activity, such as improvements in aerobic fitness, self perceptions and mood state (Weinberg & Gould, 2010). This concept also holds true for the more mature members of the female population, and with the baby boom generation reaching retirement age along with the vast improvements in medicine, this cohort is growing at a substantial rate. In the year 2010, there were 531 million people worldwide over the age of 65, and this number is due to triple by the year 2050 (Whitbourne & Whitbourne, 2011). It could therefore be assumed that the number of mature female runners is expected to increase.

Running is a natural entity, and over the course of a one mile run, an individual will make approximately 1600 foot strikes, each causing propulsive forces at to act through the lower limb (Requa, Deavilla, & Garrick, 1993; Spurgeon, 2005). Subsequently, despite documented benefits, running is known to pose one of the highest risks for the occurrence of injuries, and running related conditions; according to Requa *et al*., (1993) an average ratio of running injury occurrence to hours training existed as 1:100. Similarly, from the millions of people that are involved in running activities, 37-56% sustained an injury in the course of a one year investigation (Van Mechelen, 1992). More recently, in a systematic review of 17 reports describing the incidence of lower extremity running related conditions among runners, the overall incidence ranged from 19.4-79.3% per year (Van Gent, Siem, Middlekoop, Van Os, Bierma-Zeinstra, Koes, 2007). The most common site of lower

extremity injuries was the knee joint, attaining an average of 7.2% to 50.0% of all injuries across the studies (Van Gent et al., 2007).

3.1.2. Mature female runners and injuries.

It is widely recognised that females are at a greater risk of sustaining an overuse injury, with research results focussing on the lower extremity, including the hip, knee and ankle joints (Taunton, 2002). Several studies have investigated possible mechanisms through which these injuries occur, and it has been suggested that differences in lower limb structural kinematics, muscular characteristics, hormone conditions and gait biomechanics may be leading contenders in injury causation (See Chapter 2).

Although injury rates are prominent among the female population, a specific cohort most susceptible is that of mature female runners, as indicated by Taunton *et al*., (2002). In a prospective study, age was a significant catalyst in injury rates; an age of 50 years and above was deemed a risk factor for running-related injuries. Lanyon and colleagues supported this theory of age as a risk factor, and excluded all volunteers under 40 years when assessing prevalence of hip osteoarthritis due to changes that occur in the joints (Lanyon, Muir, Doherty, & Doherty, 2000). Therefore, with increased participation of mature females in running, understanding the effects of ageing on function and movement is increasingly important (Hageman and Blanke, 1986).

Changes in movement patterns have previously been documented among the more mature members of the population; however the majority of this research has focussed on males during walking. According to Murray, Kory and Clarkson, (1969), increased age initiated decreased stride length and cadence (increased cycle time), and increased duration of the stance phase as a percentage of the gait cycle. These changes suggest that gait alterations occur with age. Lui and Lockhart (2006) lent support to this notion of altered gait patterns with age, demonstrating significantly greater extensor moments and internal rotation at the knee joint among older walkers. Similarly, Nigg, Fisher and Ronsky (1994) suggested that as age progresses from 20 to 79 years, there is a significant reduction in range of hip flexion and extension, and knee flexion during swing phase, and from the age of forty years, rearfoot

eversion angle increases in females. However, although overuse running-related injuries are prevalent among mature females, limited evidence exists as to the differences in kinetics and kinematics among this group during running gait compared with younger female runners.

3.1.3. Biomechanics of motion and injury risk.

Within this investigation, both injuries and degenerative conditions are considered in their association with specific biomechanical variables of running gait. The knee is the weight bearing joint most commonly affected with injuries and overuse conditions, and during gait, it is often considered that abnormal biomechanics of both the proximal and distal joints will have a negative effect on the knee joint (Hunt et al., 2006). Although the pathomechanics of running related injuries and conditions are not definitive, strong relationships seemingly exist between knee conditions such as patellofemoral pain syndrome or osteoarthritis, and dynamic knee joint loading. The forces that act at the knee joint during the stance phase of gait are described in Chapter 2 (2.5.1).

3.1.3.1. Ground reaction force variables and injury risk.

As illustrated in Figure 2.13, the vertical component of the ground reaction force undergoes two peaks. This is dependant on foot position, and strike pattern during gait. Ground reaction forces are also dependent on speed of gait, and footwear characteristics, as is discussed in a later section (Bates, Osternig and Sawhill, 1983). Centre of pressure changes with foot strike were identified by Cavanagh and Lafortune (1980), who showed the location to vary from the rear third, the middle third and the front third among rearfoot, mid foot and forefoot strikers respectively. These discrepancies were also shown to affect the force trace, whereby the impact peak was most distinguishable among the rearfoot strikers, and forefoot strikers have been shown to produce smooth almost indistinguishable peaks late in stance (Lees &

McCullagh, 1984). As such, the vertical trace of the ground reaction force is used to indicate the foot strike pattern of participants during running trials within biomechanical assessments.

The vertical ground reaction force is an approximate measure of the loading of the lower extremity musculoskeletal system (Zadpoor & Nikooyan, 2011). According to Miller (1990) a justification for measuring the vertical component of ground reaction force during gait is the strong association between the magnitude of this variable and the development of overuse injuries. According to the early work of Radin and colleagues, if joints are regularly subjected to high force peaks as is expected during gait, degenerative changes take place in articular cartilage and subchondral bone (Radin, Parker, Pugh Steinberg, Paul & Rose, 1973). Similarly, Clement and colleagues have suggested that impact force peaks play a role in the development of pain and injuries among runners (Clement, Taunton, Smart & Nicol, 1981).

In a slightly later investigation, Grimston and co-workers found that a group of females who had experienced stress fractures of the lower limb exhibited significantly greater peak impact forces than a group of injury free females (Grimston, Nigg, Fisher & Ajemian, 1994). Similarly, according to Hreljac and colleagues, a comparison of biomechanical variables between a group of injured and injury free runners illustrated significant increases in both the magnitude and the rate of loading of the impact peak, suggesting that these variables may be associated with overuse injuries (Hreljac, Marshall, & Hume, 2000). In contrast, Dixon and colleagues studied a group of individuals with and without stress fractures and observed no differences in vertical ground reaction force, loading rates or the horizontal braking force (Dixon, Creaby, & Allsopp, 2006). This supported the earlier work of Crossley and colleagues, who similarly showed no difference in vertical ground reaction force or loading rate between a stress fracture and an injury free group (Crossley, Bennell, Wrigley, & Oakes, 1999). Therefore, due to the lack of unanimous results regarding the role of impact forces in the development of injury and overuse conditions, it may be considered beneficial to investigate these variables among a group of injury free, high risk individuals.

3.1.3.2. Subtalar joint motion and injury risk.

Alongside loading rates, subtalar joint pronation and knee internal rotation (Figures 2.8 and 2.11 respectively) have both been related to the development of knee injuries and debilitating conditions (Johnson & Pedowitz, 2007). At the ankle-foot complex, the subtalar joint is formed by the postcrolateral and the anteromedial joints, containing the inferior calcaneus and the talus. This joint is therefore an arthroditic joint, and a degree of congruity occurs as one portion attempts to slide over the other (Viladot, 1992). Three types of motion are present at this joint, including flexion-extension, pronation-supination and adduction and abduction (Viladot, 1992). Subtalar joint pronation is a combination of eversion, dorsiflexion and abduction. According to Shorten (2000), although all three movements are involved in pronation, rearfoot eversion is the most dominant, and therefore became a main focus of this investigation among mature female runners.

Both pronation and supination of the subtalar joint are required movements of normal human walking and running gait. Where pronation is required for shock absorption and adaptation of the foot to the ground surface, supination enables the foot to act as a rigid lever in the propulsion (push off) phase of gait to propel the runner forwards (Richards, 2008). Although the mechanisms for running related injuries are ambiguous in origin, the association between excessive rearfoot eversion and knee injury development is one link that is often reported with significant outcome. The concept that runners who over pronate (have high magnitudes of maximum rear foot eversion) are at a greater risk of sustaining an injury than those who run with a lower maximum value of rearfoot eversion, is widely held by runners and coaches alike (Hintermann and Nigg, 1998).

According to Hamill, Emmerik and Heiderschiet (1999), excessive rearfoot pronation and the timing of which, are associated with excessive load and injuries at the knee joint. Willems, and colleagues, investigated the dynamic biomechanical intrinsic risk factors of exercise related lower leg pain. In a prospective study, these researchers found increased eversion angle of the rearfoot, and accompanied increase in loading under the medial forefoot to be significantly related to the development of exercise related lower leg pain (Willems, 2006). Furthermore, Bates, James and Osternig (1978) suggested that it is often the timing of rearfoot eversion that can be a contributing factor to injury occurrence.

In contrast, Burns and colleagues showed no link between pronation and the development of an overuse injury in a retrospective study among triathletes. Interestingly a fourfold increase in risk of injury was however seen among those athletes with a supinated subtalar joint during gait (Burns, Keenan, & Redmond, 2005). Hreljac and colleagues similarly disputed the association between pronation and running injuries, and showed runners who incorporated a moderately rapid rate of rearfoot eversion to be at a reduced risk of injury (Hreljac, Marshall, & Hume, 2000). It is however noted that this retrospective investigation involved a comparison of injury free runners, and those who had sustained an injury in the previous year but were free from injury at the time of testing. Therefore at the time of testing, both groups were free from injury. Additionally, no significant difference was noted in the magnitude of peak rearfoot eversion angle between the two groups, and the mean values of 4.0^{o} and 5.9^{o} for the previously injured and injury free groups were within normal ranges (Hreljac et al., 2000).

During normal gait, maximum eversion, tibial internal rotation, and knee flexion all occur during the same phase of stance. This relationship has been described by Seigler and colleagues as kinematic coupling, whereby differences in the timing of pronation of knee flexion result in contradictory rotations of the tibia and knee joint (Seigler, Chen, & Schneck, 1988). However if the subtalar joint continued to pronate as the knee extended, the tibia would be subject to antagonistic counter rotations. Therefore, delayed or prolonged maximum rear foot eversion could lead to excessive stress at the knee joint, increasing vulnerability to injury.

It has been proposed that the link between rearfoot eversion and injuries is therefore attributed to the coupling between these two joints. As described above, during heel strike of running gait the distal portion of the calcaneus touches the ground and everts. As the talus and the tibia are both contained within the ankle joint, and the talus is connected to the calcaneas through the talocalcaneal interosseous ligament, movement of the calcaneum is transmitted to the tibia (Jenkins, 1998). Therefore, at the distal end of the tibia, rotation of the foot about the longitudinal axis is transmitted to the tibia, imposing a rotation about its long axis. Thus, rotation of the foot from a supinated to a pronated position results in an internal rotation of the tibia about its longitudinal axis (Stergiou, Bates & James, 1999). This notion supports Hintermann and Nigg (1998) who stated that calcaneus eversion is linked through the talus to tibial internal rotation. Furthermore, according to Olson (2007), increases in the

magnitude, velocity and acceleration of subtalar joint eversion result in a similar increase in the magnitude, velocity and acceleration of tibial internal rotation. This in turn is then transmitted to the knee to cause an increase in the internal rotation and adduction of the knee joint. During the stance phase of gait, it is not only the magnitude of rear foot eversion, but the means by which it is transferred into tibial internal rotation, that may be crucial to the overloading and stress applied to the knee joint (Eslami, Begon, Farahpour, Allard, 2007). This is supports Hintermann and Nigg (1998), who suggested the transfer of the eversion movement of the calcaneus to the tibia may be associated with overloading of the knee joint. Levinger and Gilleard (2005) similarly showed a correlation between the time of peak eversion and tibial rotation among a group of walkers with patellofemoral pain syndrome. Results indicated a delayed and prolonged rearfoot eversion among the patellofemoral pain group, which was not however seen among the injury free group. It was therefore suggested that prolonged rearfoot eversion reduced the time of tibial rotation, which may affect the normal patellofemoral mechanics during gait (Levinger & Gilleard, 2005). Therefore, when investigating mechanisms of injury to the knee joint, it is deemed important to assess the magnitude, timing and velocity of rear foot eversion, as well as the similar variables associated with tibial internal rotation.

Although the term 'excessive' has not been clearly defined, and the aetiology of lower limb injuries is certainly multi-factorial, high values of pronation may disrupt normal biomechanics during gait, and the weight of the literature presented above suggests a relationship between excessive rear foot eversion and knee injuries.

3.1.3.3. Sagittal plane ankle motion and injury risk.

The range of ankle plantar flexion is greater than that of dorsiflexion, with average values for running gait between 30-50° and 20-30° of the former and latter respectively (Norris, 2004). However these values often vary between individuals and it has been frequently suggested that ankle dorsiflexion range can be a strong predictor of injuries to the lower limb (Pope, Herbert, & Kirwan, 1998). Among heel strikers, long distance running involving a heel-toe

motion, works the ankle through its full range (Grisogono, 1984). Limitations to the available range of motion are likely to impede gait and increase the risk of injury.

Ankle dorsiflexion range and ankle plantar flexion strength are important components of gait (Drewes, McKeon, Kerrigan, & Hertel, 2009). Malliaras, Cook and Kent (2006) hypothesised that impairment of these variables may increase the load applied to the patella tendon, and result in injury (patellar tendinopathy). Their cross sectional study investigated the association between range of motion and the presence of injury among volleyball players. Results suggested that patellar tendinopathy was significantly associated with reduced range of ankle dorsiflexion, when compared with normal tendons. Furthermore, less than 45° ankle dorsiflexion range increased the risk of this injury by 1.8-2.8 times (Malliaras et al., 2006). It is noted that this study investigated range of motion during jumping rather than over ground running, however the association between dorsiflexion and injury still bears importance.

Drewes, McKeon, Kerrigan, and Hetrel (2008) later supported this theory, suggesting that during gait, limited ankle dorsiflexion range may increase the risk of ankle sprains and subsequent recurrent injuries. Reduced terminal dorsiflexion prevents the ankle from reaching its stable closed-packed position at midstance during dynamic tasks such as running. Additionally, reduced dorsiflexion may increase the vulnerability of the ankle to hypersupination where the ankle inverts excessively and the medial longitudinal arch is lifted from the ground; another risk factor for injury (Lindsay, 2008). Reduced range of motion at the ankle may be a consequence of reduced flexibility in the muscles, a common factor associated with increasing age. As well as investigating range of motion at the joint, the eccentric action of the calf muscles that controls dorsiflexion, and its association to injury has also been investigated. Naicker, McClean, Esterhuizen and Peters-Futre (2007) reported low peak dorsiflexion torque as a factor associated with ankle injury among female hockey players, consistent with the earlier findings of Baumhauer, Alosa, Renstrom, Trevino and Beynnon, (1995) who found a greater plantar flexion to dorsiflexion strength ratio was associated with a greater incidence of injuries among 145 college-aged athletes.

Inadequate dorsiflexion control may be a result of weakness in the anterior tibial muscles, or the incidence of these muscles being overpowered by spasticity of the triceps surae muscles (Whittle, 2003). The weight of the research supports the notion that altered ankle biomechanics play a role in the genesis of running related lower limb injuries, and therefore

needs to be assessed when determining the cause or potential management of such conditions (Willems, De Clercq, Delbaere, Vanderstraeten, De Cock, & Witvrouw, 2006).

3.1.3.4. Knee internal rotation and injury risk.

As previously described (section 2.4.3) the rearfoot and the knee are mechanically linked by the tibia, and owing to the inclined axis of the subtalar joint in the sagittal plane, eversion of the foot leads to internal rotation of the knee (Willems et al., 2006). As rotation of the foot occurs with subtalar pronation, this motion is transmitted to the tibia through its longitudinal axis (Stergiou et al., 1999). Although a small degree of rotation is inherent during gait, excessive amounts have been related to injury development particularly at the knee joint. At the knee joint, altered tibiofemoral rotation during the stance phase of gait is proposed to alter the normal patella tracking and increase patellofemoral joint compression in specific areas (Norris, 2004). According to Olson (2007), high values of internal rotation of the knee interferes with normal knee function as the patella is shifted medially, which in turn increases the vertical pull of the vastas medialis obliques, increasing the compressive forces acting on the knee joint. Additionally, internal rotation causes decreased patellofemoral contact area, which in turn increases the pressure in specific areas of the knee, increasing the wear on the medial condyles of the knee joint (Norris, 2004), and according to Olson (2007), internal rotation of the knee joint causes excessive overstretching or twisting of the ligaments or tendons or increased shearing forces on the bursa surrounding the knee. In addition, if the timing between rearfoot eversion and knee internal rotation is disrupted, the normal coupling between the joints is affected and an antagonistic relationship occurs. This in turn could then lead to soft tissue stress, and with multiple repetitions as would be expected during running, an injury is likely to occur (Stergiou et al., 1999). Common knee conditions related to such mechanisms include iliotibial band syndrome as the iliotibial band attempts to restrict the excessive rotation of the tibia (Noehren, Davis, & Hamill, 2007). Additionally, patellofemoral pain syndrome can occur with excessive internal rotation of the tibia, as this can create lateral mis-tracking of the patella in the patella grove of the femur (Olson, 2007).

3.1.3.5. Q-angle and injury risk.

The Q angle is a measure of the alignment between the femur and the tibia in the frontal plane (McGinnis, 2005). The patella is the largest sesamoid bone in the body, and is attached to the quadriceps tendon above and the patella tendon below which inserts in to the tibial tubercle (Norris, 2004). The vastus medialis, vastus intermedius, vastus lateralis and rectus femoris are the four main elements of the quadriceps muscle which combine to form the quadriceps tendon. This surrounds the patella and continues beyond it to become the patellar tendon (Whittle, 2003). The direction of pull of the quadriceps tendon is along the shaft of the femur, and that of the patella tendon is almost vertical; therefore a frontal plane angle is formed between these two lines (Norris, 2004). Specifically the angle is created in the intersection of two lines; the line from the anterior superior iliac spine to the centre of the patella, and the line connecting the centre of the patella to the tibial tuberosity (Figure 2.4) (Heiderscheit, Hamill, & Caldwell, 2000).

The Q angle is routinely implemented to assess anatomical alignment of the lower extremity in regard to the patella (Sweden, 2001). In the clinical situation, malalignment has been speculated as a risk factor for running injuries (Caine, Harmer & Schiff, 2009). During activities such as running, the knee joint is exposed to high levels of stress due to the imposed ground reaction force and the subsequent tension in the muscles crossing the joint. In normal alignment, the resultant force ideally passes through the joint centre of the knee, minimising the risk of injury occurrence. Normal values for the Q angle are in the region of 10° to 15°; according to the American Orthopaedic Association (AOA, 2011) Q angles reaching a value above 15° are defined as “excessive” (Heiderscheit et al., 2000; Norris, 2004). High Q angles can be a result of anatomical abnormalities such as the displacement of the tibial tubercles to the outer side rather than the centre of the shin bone, or if there is an obvious valgus (knock-kneed) position during standing (Grisogono, 1984). This was supported by Mizuno and colleagues who showed a reduction in Q angle to reduce the valgus angle at the knee, and reduce the lateral tibiofemoral contact pressure (Mizuno, Kumagai, Mattessich, Elias, Ramrattan, Cosgarea, & Chao, 2001). Additionally, high Q angles have often been related to increased pronation at standing or during the stance phase of gait (Grisogono, 1984).

As stated, greater Q angles produce malalignment of the lower extremity. An early study by Subotnick (1975) suggested that increased Q angle causes greater rearfoot eversion, and

subsequent pronation, as a result of knee valgus. During running gait, where high forces are attenuated through the lower extremity, large Q angles can therefore place the body at a greater risk for injury occurrence. According to Grisogono, (1984), a large angle between the patellar tendon and the quadriceps muscle causes the patella to deviate from its normal pathway on the distal end of the thigh bone. During a repetitive movement such as running, these mechanical factors may cause persistent minor dislocations of the patellar. This in turn could lead to joint laxity of the medial ligaments, sub-patella swelling, patella tracking problems, and ultimately disruption of the patellofemoral joint (Grisogono, 1984).

This theory was supported by Sweden (2001) who suggested that high Q angles cause a valgus force to act on the patella, and can predispose the knee to lateral patella subluxation or dislocation, increasing the vulnerability of the joint. Furthermore, the notion of a link between high Q angle and patellofemoral injuries was illustrated by Messier, Davis, Curl, Lowery and Pack (1991), who suggested that a relationship exists between certain anthropometric and biomechanical measures, and patellofemoral pain. Results of their investigation showed Q angle to be a significant discriminator between patients presenting patellofemoral pain, and controls, lending support to the suggestion that greater values of Q angle may be a contributing factor to the aforementioned condition. Similarly, Rauh, Koepsell, Rivara, Margherita and Rice (2006) investigated Q angles among high school athletes, and found boys with a Q angle above 15 ^{o} and girls above 20° were twice as likely to incur an injury during cross country running. This was supported in a subsequent prospective study, which indicated runners with a Q angle above 20 ^{o} had a risk of injury to the knee joint 1.7 times that of runners with an angle below 15° (Rauh, Koespell, Rivara, Rice and Margherita (2007). According to Heiderscheit et al., (2000), the association between rearfoot eversion and excessive Q angles as supported by Subnotnick, 1975, may subsequently result in delayed maximum eversion angle and tibial internal rotation angle. As such, increased torsional load is applied to the tibiofemoral joint as the knee attempts to regain its normal extension phase. Therefore, in light of the literature presented, it is suggested that high Q angles may be a contributing factors to injuries occurring at the knee joint during running. It is therefore important to assess this variable when reviewing possible injury mechanisms in a group of injury prone individuals.

3.1.3.6. The knee external adductor moment.

Alongside kinematic variables, a common clinical measure of gait when assessing knee conditions is the knee external adductor moment; a feature consistently associated with medial knee loading and osteoarthritis development (Lui & Lockhart, 2006). As described, joint moments are a product of the force acting through the joint, and the distance of the force from the joint centre (moment arm). Both the ground reaction force component and the moment arm are independent variables that can each be manipulated separately, and have an individual influence on the calculated moment (Hunt, Birmingham, Griffin, & Jenkyn, 2006). As such, an increase in the magnitude of ground reaction force, or malalignment of a joint, could both contribute to increases in joint moments (Richards, 2008). The relative influence of the magnitude of the ground reaction force and the size of moment arm on the moment calculated at the knee joint was investigated by Hunt et al (2006). Here it was shown that the calculated moment was more greatly associated with the length of the moment arm than with the magnitude of the ground reaction force variable (Hunt et al., 2006). Similarly, Stief and colleagues later suggested that higher moments at the knee joint were considered a product of malalignment as opposed to an increase in magnitude of ground reaction force (Steif, Bohm, Schwirtz, Dussa, & Doderlein, 2011).

As well as suggesting a greater influence of moment arm length, both aforementioned studies have also shown an association between greater moments and the development of injuries or overuse conditions. Hunt and colleagues found significantly greater knee external adductor moments among a group of osteoarthritis patients compared with a group of asymptomatic individuals (Hunt et al., 2006). This was later supported by Davies-Tuck and colleagues, who found the presence of medial mensical tears to be positively associated with an increase in the peak knee external adductor moment among post menopausal women (Davies-Tuck, Wluka, Teichtahl, Martel-Pelletier, Pelletier, Jones, Ding, Davis, & Cicuttini, 2008). Steif and colleagues showed an increase in knee adduction moment, and knee internal rotation moments among a group of young patients with cartilage degeneration compared with a symptom free group (Steif et al., 2011). According to Stefanshyn and colleagues, loading of the knee joint represented by calculation of the moments, may also play a functional role in the onset of patellofemoral pain syndrome among runners (Stefanyshyn, Stergiou, Lun, Meeuwisse, & Nigg, 1999). Results from their investigation illustrated significantly higher

frontal plane and transverse plane moments among a group of injured compared to injury free runners, suggesting that increased moments may be a contributing factor to the onset of this condition (Stefanyshyn et al., 1999). Although research has looked into the association between the knee adductor moment and osteoarthritis, it is considered that the weight of this research is based on symptomatic patients during walking trials. Within this thesis, a focus will remain on running trials, to further investigate this association among runners.

The above research has highlighted the link between joint moments and injury, and expressed the importance of considering the independent ground reaction force and moment arm components individually when investigating moments as a potential mechanism for injury and overuse conditions.

Increasing age is known to cause structural changes within the female body, specifically from the age of 40 years and above (Lanyon, Muir, Doherty, & Doherty, 2000). These structural changes may contribute to potential differences in the running gait patterns between young and mature females. An understanding of the dynamic functional anatomy is of critical importance to appreciate the pathomechanics of injuries and conditions such as osteoarthritis (Olson, 2007).

3.1.4. Study One intentions and hypotheses.

The purpose of this study was to compare the running gait of young and mature females to investigate whether mature females demonstrate gait characteristics that may predispose them to the development of running related knee injuries and overuse conditions. One research hypothesis was assessed:

1. When compared to a younger group, mature female runners will exhibit greater values of peak rate of loading (vertical ground reaction force), peak knee internal rotation, peak rearfoot eversion, and peak knee external adductor moment.

3.2. Methods.

3.2.1. Participation selection.

Thirty female participants (15 mature and 15 young) were recruited on a volunteer basis through both the local Women's Running Network and the University of Exeter Exercise and Sports Science cohort. A letter was sent to the running network, refreshing previous alliance and detailing the proposed study. Additionally, a similar email was forwarded to the younger female students. Participants were included based on age, the younger group aged 18 to 25 years (mean 20.9 years, sd 1.9 years) and the mature group aged 40 to 60 years (mean 49.3 years, sd 4.6 years), as 40 years was classified as the age at which structural changes begin to occur at the joints (Lanyon et al., 2000). All females had a minimum of 12 months running experience, and were participating in a minimum of three one hour sessions of running per week. 10 km run-time was used to indicate running standard (Table 3.2). Prior to gait testing, each subject was requested to complete a PAR-Q health and safety form to quantify relative ages, heights, mass, running experience and overall health status to ensure participants were eligible for participation. All females were requested to state their experience of overuse injuries, and the use of orthotics. Although not a study requirement, it was noted that no orthotics were used or injuries presented by any participant. Participants also filled out the KOOS (Knee Osteoarthritis Observation Survey) questionnaire as advocated by Roos and Toksvig-Larsen, (2003), to assess the prevalence of or potential for knee osteoarthritis development (Appendix B). Although unexpected in the younger age group, all subjects were required to complete the form. A summary table of subject information is presented at beginning of the results section (Table 3.2). The majority of previous studies have used the WOMAC scale when assessing osteoarthritis presence and severity, however recently the KOOS score has been shown to improve validity of results and deemed at least as responsive as the former (Roos and Toksvig-Larsen, 2003). Therefore in light of the described research, the KOOS score was considered most appropriate for the present study.

In order to assess the potential or presence of osteoarthritis, the survey recorded scores on five subscales; Pain, Other Symptoms (Symptoms), Function in Daily Living (ADL), Function in Sport and Recreation (Sport/Rec), and Knee Related Quality of Life (QOL).

Reviewing the results of the five Likert boxes then indicated the prevalence of osteoarthritis; total scores ranged from 0 to 100 representing extreme to no symptoms respectively.

Ethical consideration was given to all aspects of the proposed research study. It was emphasised that all participants were above the age of 18 years, the procedure was entirely non invasive, and that all participants had the right to withdraw at any moment in time. A proposal was handed to the ethical committee at the School of Sport and Health Sciences, who approved the experimental protocol and enabled a solid foundation upon which to carry out the intended research.

3.2.2. Motion analysis and data capture.

Subjects were required to wear their usual running attire including their own running trainers, to encourage them to run in their usual style. Three-dimensional kinematic analysis of the right lower limb was performed using an eight camera motion capture system (120Hz, Automatic, opto-electronic system; Peak Performance Technologies, Inc., Englewood, CO.), synchronised with a single floor mounted force platform containing strain gauge transducers (960Hz, AMTI, Advanced Mechanical Technology, Inc., Massachussetts). Data were synchronized within the Vicon software using initial foot strike as an automatic event detection (vertical force > 10N). The eight cameras were positioned in an oval shape focussing on the force plate. A static L-shaped frame denoting the horizontal axis of motion, and calibration wand of length 0.98 m with two reflective markers of known distance apart were used to calibrate the system.

The spatial movement of the lower limb was determined from the position of spherical markers. Eleven reflective markers were attached to the greater trochanter, medial and lateral knee at the tibial plateaus, the musculo tendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon, the mid tibia below the belly of the tibialis anterior, the superior and inferior calcaneus, the lateral malleolus, the third proximal head of the third metatarsal, and the distal head of the fifth metatarsal joint to denote the anatomical position of the thigh, shank and foot (Soutas-Little et al., 1987) (Figure 3.1). This was based on the model designed by Soutas Little, Beavis, Vertraete, & Markus, (1987), who used 11

reflective markers to define the anatomical co-ordinate system of the lower limb and track segment motion during gait. Within this model, each body segment was modelled as a rigid body, maintaining a constant distance between two points. Kinematic data were then filtered with a quintic spline processor (peak performance default optimal smoothing technique using 5th degree quintic polynomials; Woltring, 1985); the spline being a smooth curve passing through a minimum of two points. The error was determined using a default method within Peak Vicon Technologies, enabling raw coordinate data to be smoothed based on this estimate of error in relation to a central trend (Wood, 1982).

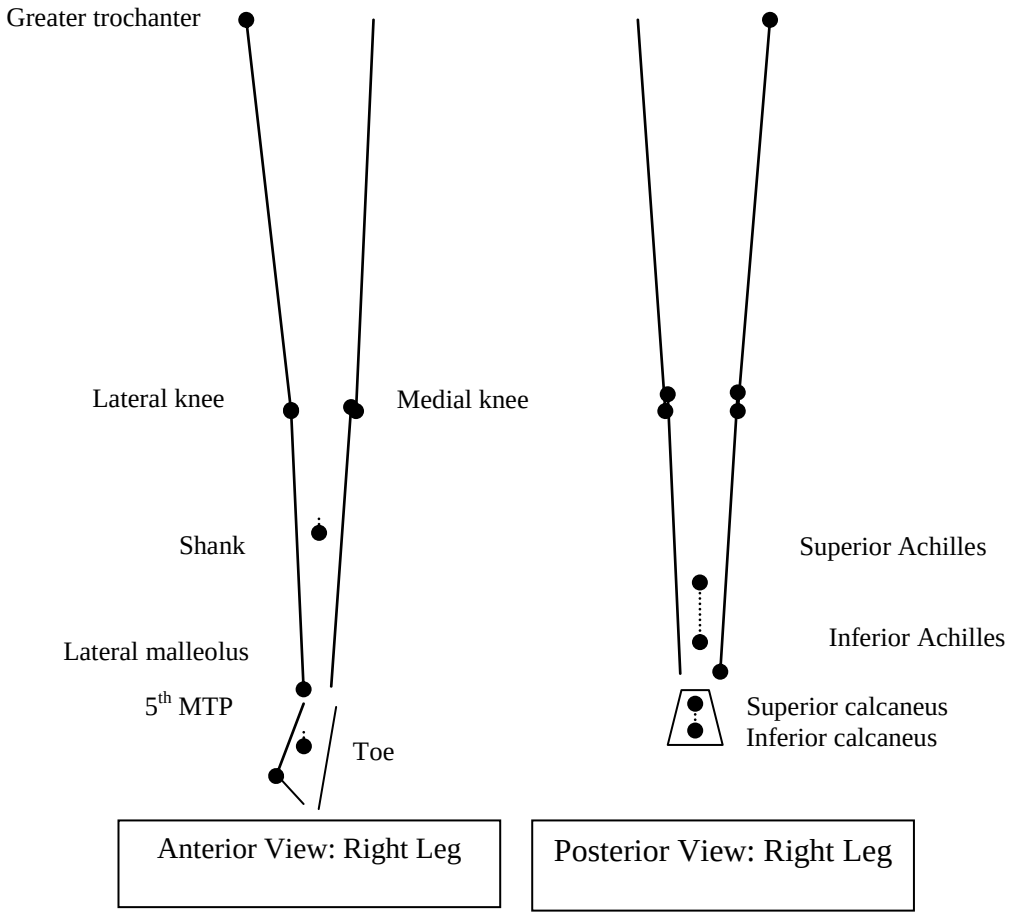


Figure 3.1. Annotated illustration of marker placement along the right lower limb; both anterior and posterior views. Not to scale.

Following the laboratory set up phase, subjects were given an adequate habituation period. Each female was permitted to self select the approach distance to the force plate, within the boundaries of the laboratory, and provided feedback regarding the running speed. Running velocity was recorded using two infrared timing gates, situated 2 metres apart, categorising each trial as acceptable at 3.5 $m.s^{-1}$ ± 5%.

All kinematic data were referenced to a relaxed standing position, where by neutral joint angle data were obtained to provide anatomically relevant values. Each participant performed a single standing trial on the force platform, enabling recognition of individual marker location and body mass. These data were also used to compare standing Q angles between females, and then subtracted from the kinematic values obtained during the running trials, to calculate body position during each trial (Hunt *et al*., 2006). This method enabled running data to be compared between individuals, without the influence of possible differences in standing position.

Following the familiarisation period and collection of neutral data, the force plate was zeroed and each individual performed ten running trials, ensuring complete right foot contact with the platform on each occasion. Data were collected for one second within which the force plate was contacted. Running trials where stride adjustment occurred prior to force plate contact were repeated. Trials were deemed acceptable when running gait remained visibly natural, complete right foot force plate contact occurred, running velocity fell within 5% of the targeted speed, and markers remained visible throughout the trial. Each female was then permitted a 30 second rest period to reduce the possibility of fatigue, the force plate was zeroed, and the next trial commenced.

Ground reaction force data were sampled at a frequency of 960 Hz. The vertical force component applied to the plate during ground contact was the main point of interest within the ground reaction force data. Although not a requirement for this investigation, during the familiarisation period, all females were categorised as heel strikers. As such, each force time graph highlighted impact force peak, and the rate of loading denoted by the gradient during the impact phase. Details of the method used to calculate loading rate are described in Chapter 2 (2.5.1). Figure 3.2 shows a sample vertical force time history, highlighting important aspects including loading rate.

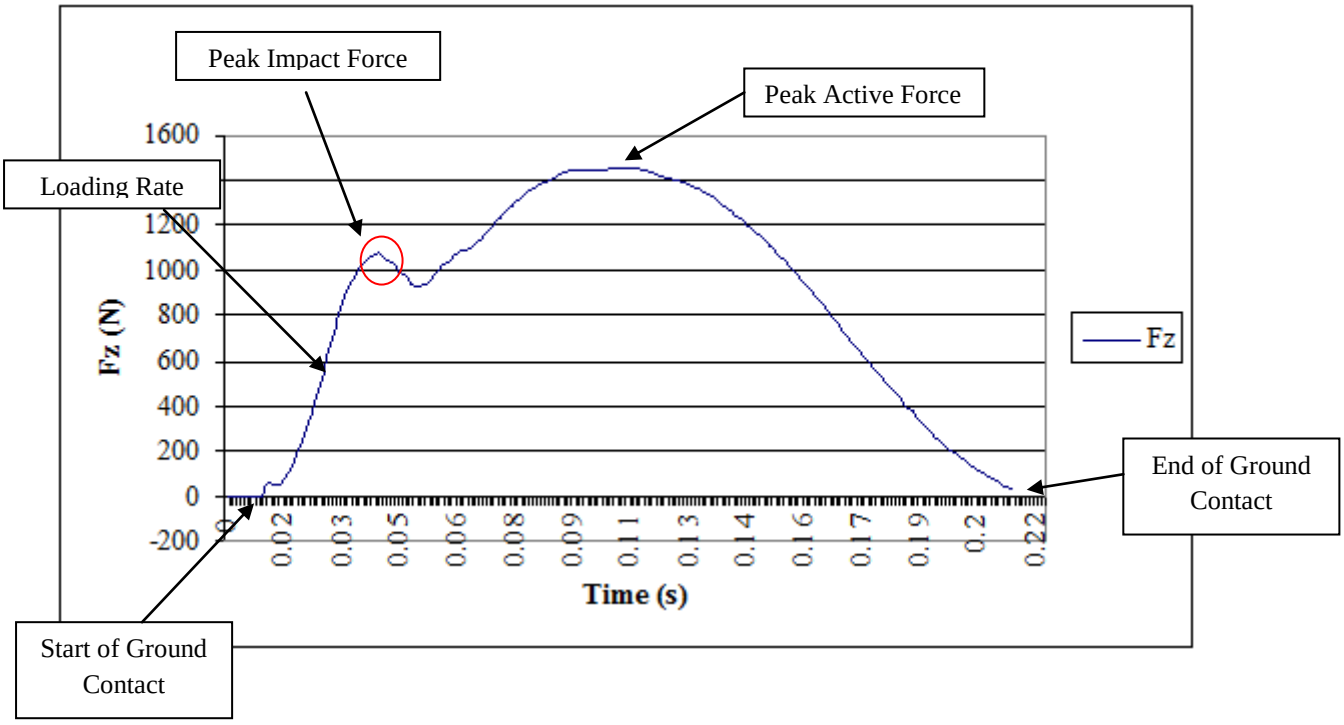


Figure 3.2. Sample vertical force time history. Data taken from subject 19, trial 5.

Three-dimensional analysis of each subject's running gait occurred through synchronisation of the eight cameras situated in the laboratory. Kinematic data were sampled at a rate of 120Hz. Peak angles including peak knee internal rotation and rearfoot eversion were determined and compared between subjects. Finally the transformed co-ordinates and ground reaction force data were exported from Peak into MATLAB (Version 7.0, Mathworks, Inc) which was used to determine frontal plane knee moments; the knee external adductor moment. Moments were then normalised for body mass, and presented in Nm/kg, enabling between subject comparisons.

3.2.3. Kinematic and kinetic variables.

The Q angle was calculated for each participant during the standing static trial. Two seconds of data were captured at 120 Hz, and an average value was calculated. The Q angle was measured as the angle between the line from the medio-lateral coordinate (x_{troch}) of the marker on the greater trochanter to the central patella measured between the two x-coordinates of the markers on the knee, and the vertical line from the central patella to the tibial tubercle (Kirtley, 2006). In this investigation, the raw value was measured as an absolute angle, and the relative Q angle was calculated from the following:

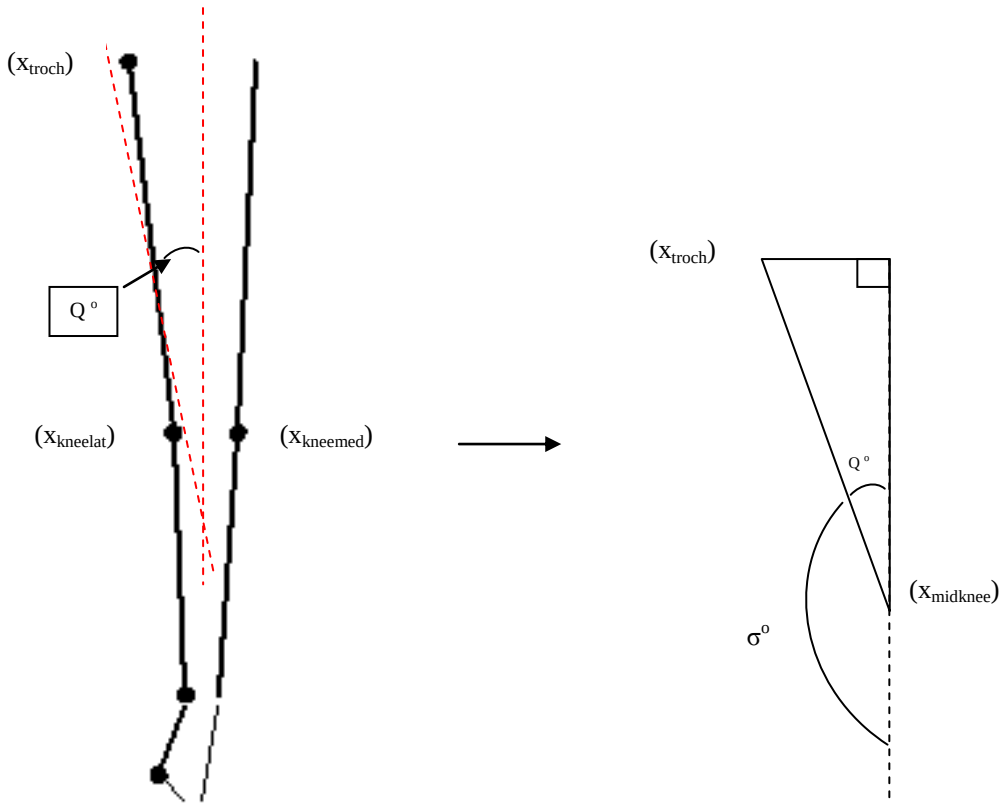


Figure 3.3. Quadriceps angle calculation.

$$x_{midknee} = x_{kneelat} - x_{kneemed}$$

Equation 3.1

$$Q^o = 180^o - \text{Calculated absolute angle } \sigma^o$$

Equation 3.2.

The following variables were assessed to characterise each individuals running gait, and attain group means. Each angle was measured as the peak value during stance. A detailed description of angle calculations is provided in Chapter 2. A summary of the variables measured in this investigation is displayed below (Table 3.1). Statistical analysis of obtained results involved a series of independent sample t-tests to identify significant differences between the two groups for each variable measured.

Table 3.1. Summary table illustrating the specific running gait variables for analysis.

Gait Parameter		**Description**
Kinetic Variables	Peak Impact Force (Fz)	Maximum value achieved in the first peak of the vertical component of the ground reaction force trace.
	Peak Loading Rate	Gradient of vertical force time curve. The rate at which the impact force occurs within first 50ms of force generation.
Kinematic Variables	Peak Ankle Eversion Angle	Angle created between the line through the two points on the calcaneus, and those on the inferior and superior Achilles tendon, measured in the XZ reference plane.
	Peak Knee Internal Rotation	Rotation of the tibia relative to the femur about a longitudinal axis embedded within the tibia. When the frontal plane of the tibia is parallel to the medial-lateral axis of the femur knee rotation is measured as 0°.
	Peak Dorsiflexion Angle	Angle created between the foot and the tibia, calculated in the sagittal plane. Neutral dorsiflexion angle is 90°.
Joint Moments	Frontal Plane: Peak Knee Adduction Moment	Moment produced as the net ground reaction force passes medially to the knee; a combination of the frontal plane force magnitude and the moment arm to knee joint centre.

3.2.3.1. Joint moment calculation.

Determination of the joint centre location and moments is described in detail in chapter 2 (2.4 and 2.6). Prior to initiation of data collection, two anatomical measurements were taken with the consent of each participant. Width measurements of the forefoot (widest as observed in shoe) and ankle (from lateral to medial malleolus bones) were taken using a set of callipers. Based on the cadaveric studies by Dempster (1955), the ankle joint centre is considered to fall within the body of the talus in the axis that passes through the line between the centre of the lateral and medial malleoli (Nair, Gibbs, Arnold, Abboud, & Wang, 2010). As illustrated above (Figure 3.1), a marker is placed on the lateral malleolus; therefore the ankle joint centre is calculated using the position of this marker and the measured diameter to the medial malleous. The diameter of the forefoot enabled the foot to be measured as a rigid body, repressed as the centre line through the lower calcaneus marker and the mid point of the forefoot. The foot centre of mass used for the moment calculation was then determined using the data from Dempster (1955).

3.3 Results.

3.3.1. Participant information and analyses of KOOS assessment.

Table 3.2 presents a summary of participant information. Included are the results from the KOOS survey, which revealed all subjects both mature and young to be categorised as low risk in all subscales; pain, symptoms, activities of daily living, sport/recreation, and quality of life, with an average score of 96.5 and 90.8 out of 100 for the young and mature respectively (Table 3.2).

Table 3.2. Presentation of mean group subject information, including KOOS scores.

		YOUNG	**MATURE**
AGE (years)		20.9 (1.9)	49.3 (4.6)
HEIGHT (cm)		165.5	159.3
WEIGHT (kg)		60.9 (1.3)	58.2 (2.4)
10k Time (minutes)		55 mins	57 mins
KOOS	***Pain***	97.9 (1.6)	94.2 (2.1)
	Symptom	98.1 (1.5)	91.5 (3.5)
	Activities of daily living	98.7 (1.4)	92.9 (2.04)
	Sport & Recreation	93.8 (4.2)	87.8 (2.5)
	Quality of life	94.1 (4.04)	87.9 (3.4)
	MEAN	**96.5 ± 2.4**	**90.8 ± 2.9**

A sample line graph illustrating group means for each subscale of the KOOS survey is presented in Figure 3.4. Results from a one-way ANOVA for the two sets of results illustrated a significant difference in the scores for the two groups ($p = 0.0089$, $p<0.05$) (Table 3.3).

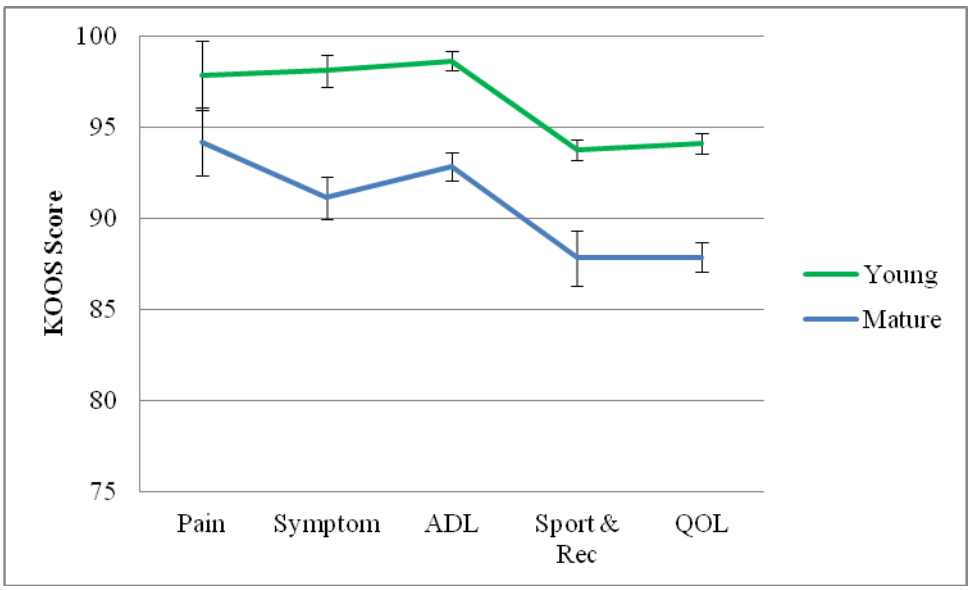


Figure 3.4. Line graph illustrating average KOOS scores.

Table 3.3. Results from a one-way ANOVA for group mean KOOS scores.

ANOVA	**Average**	**Variance about the mean**
Mature Females	90.8	8.37
Young Females	96.5	5.52
Source of Variation		
	Degrees of Freedom	P-value
Between Groups	1	0.0089*

Results from a series of regression tests are presented in Table 3.4; illustrating the relationship of each variable with age of the mature females. Individual data illustrating these relationships among the mature females are presented in Figure 3.5.

Table 3.4. Statistical output from regression tests between KOOS score for each subscale and age.

	Regression Analyses	
	R value	P-value
Age: Pain	-0.612	**0.015***
Age: Symptoms	-0.55	**0.033***
Age: ADL	-0.57	**0.026***
Age: Sport & Rec	-0.58	**0.023***
Age: QOL	-0.58	**0.025***

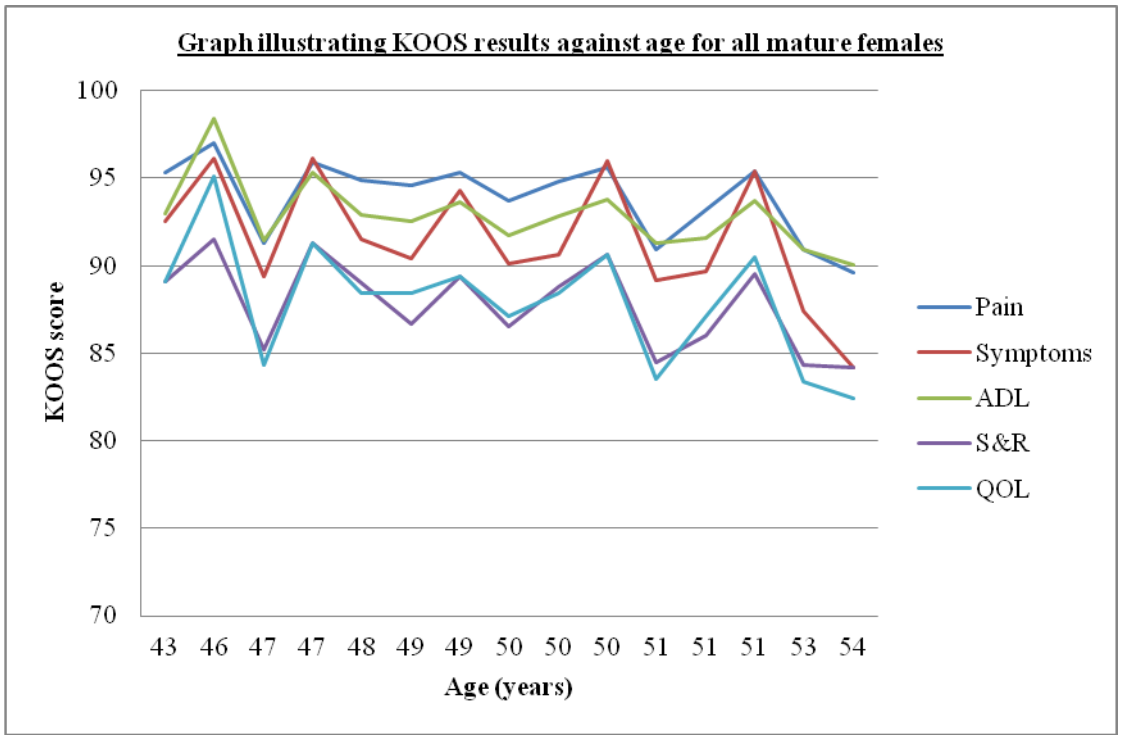


Figure 3.5. Graphical representation of the association between individual KOOS scores and age (mature females).

3.3.2. Assessment of biomechanical variables during running gait.

Individual mean data for each biomechanical variable is presented in the appendix (Appendix C). Group averages are summarised below, highlighting where relevant, significant differences between mature and young female runners.

Table 3.5. Table displaying kinematic and kinetic group means for each biomechanical variable assessed during gait.

		Mean results acquired from 10 trials.		
		MATURE (n=15)	**YOUNG (n=15)**	**p value**
Average Q Angle (Degrees)	Mean	**8.03**	**6.62**	0.57
	SD	7.3	6.3	
Peak Impact Force (BW)	Mean	**2.0**	**1.9**	0.49
	SD	0.6	0.5	
Peak Loading Rate ($BW.s^{-1}$)	Mean	**54.9**	**36.5**	0.003*
	SD	9.02	8.92	
Ankle Eversion Angle (Degrees)	Mean	**11.1**	**6.7**	0.00029*
	SD	3.2	2.57	
Ankle Dorsiflexion Angle (Degrees)	Mean	**23.04**	**27.64**	0.00056*
	SD	3.01	3.44	
Knee Internal Rotation (Degrees)	Mean	**12.5**	**8.3**	0.00007*
	SD	2.3	2.57	

3.2.2.1. Standing Q angle assessment.

Individual Q angle data is presented in Appendix C (C1). As only one neutral trial was taken, standard deviations are not presented for each individual. Group mean data is displayed graphically in Figure 3.6. Analysis of the group mean data showed the mature females to demonstrate a q angle that was 1.4^{o} higher than that of the younger females. However results from the independent sample t test showed this between group difference to be non significant (p = 0.56).

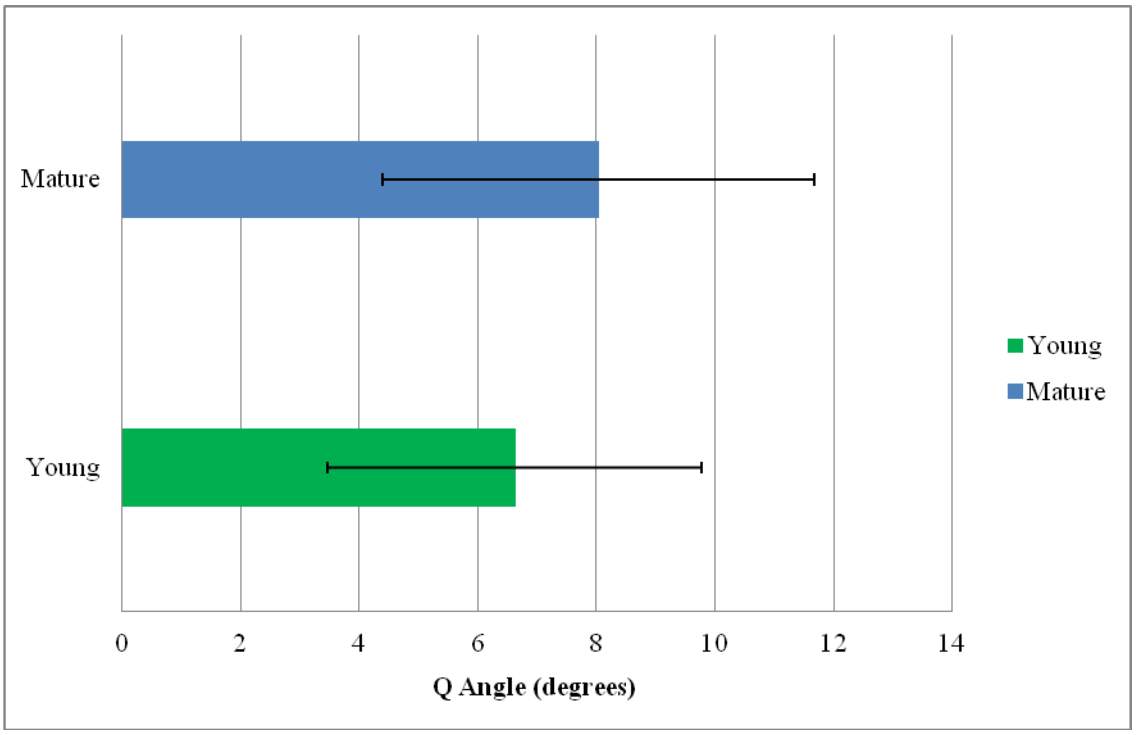


Figure 3.6. Bar chart illustrating mean Q angle data for the two groups, including standard deviations.

3.2.2.2. Ground reaction force variables.

Figure 3.7 shows a sample vertical force time graph illustrating results from 9 trials performed by Subject 19 (young). One trial was omitted due to an error in the fore plate measurement. Each curve exhibited two distinct peaks; peak impact force and peak active force, a characteristic pattern indicative of a heel-toe running gait. Here it was noted that all participants employed this same running style. As indicated in Table 3.5, no significant difference was observed in peak Fz, however a significant difference was found in peak loading rate at the $p<0.05$ level ($p = 0.49$, $p = 0.003$). Differences between the group means for both variables are presented in graphical form in Figure 3.8. For clarity of the display, standard deviations were not included in this graph. Among the mature females, a low and non significant correlation was found between KOOS score and loading rate ($r = -0.06$, $p = 0.0412$).

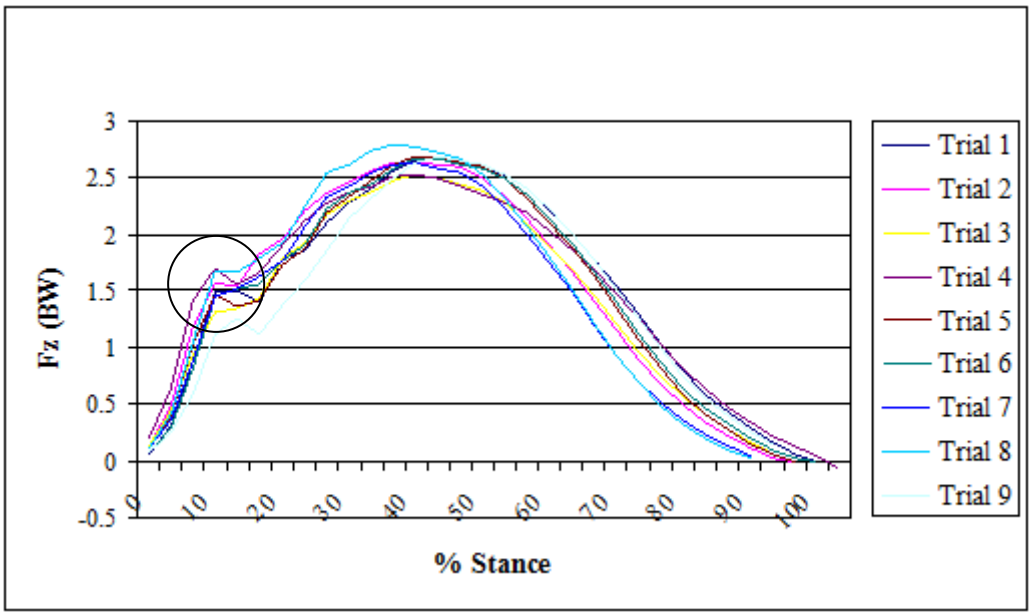


Figure 3.7. Sample vertical force ground reaction time history produced by Subject 19 (young) (FzBW).

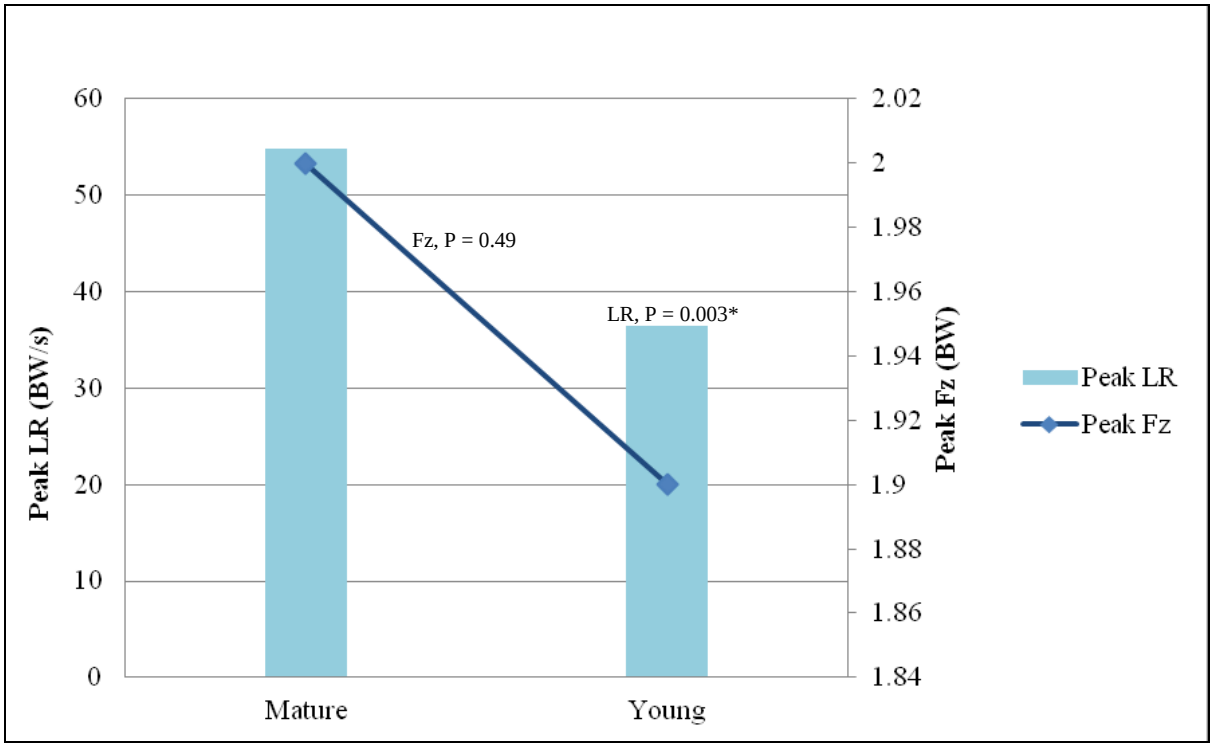


Figure 3.8. Sample bar graph illustrating mean peak vertical force (Fz), peak loading rate (LR), and significance (p) values for both groups. Data scaled to body weight.

3.2.2.3. Rearfoot eversion angle.

Figure 3.9 illustrates a sample angle time history produced by subject 14 throughout the ten running trials (mature). Data from two trials was omitted due to incomplete force plate contact. As illustrated in the graph, the rearfoot moves from a slightly inverted position at heel strike to peak eversion at early to mid stance. The rearfoot then returns to inversion as ground contact progresses through to late stance and just prior to toe off. Calculation of the group mean values illustrated an overall difference of 5.6 degrees between mature and young female runners, which was significant at the $p < 0.05$ level (Table 3.5). This difference is highlighted in the bar graph (Figure 3.10).

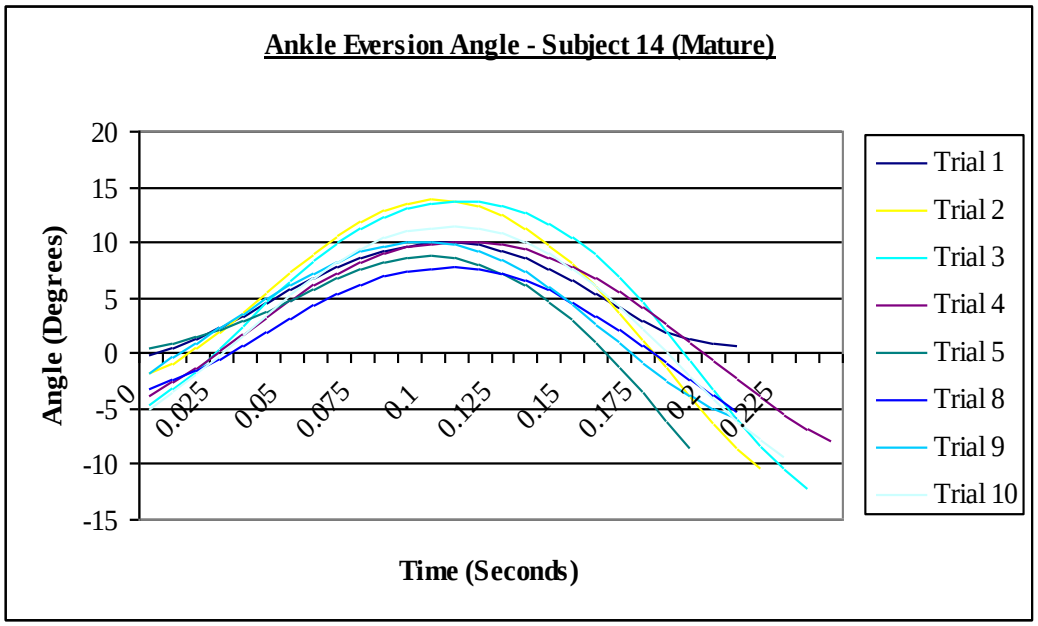


Figure 3.9. Graphical representation of the ankle inversion/eversion angle during ground contact for subject 14 performing 10 running trials.

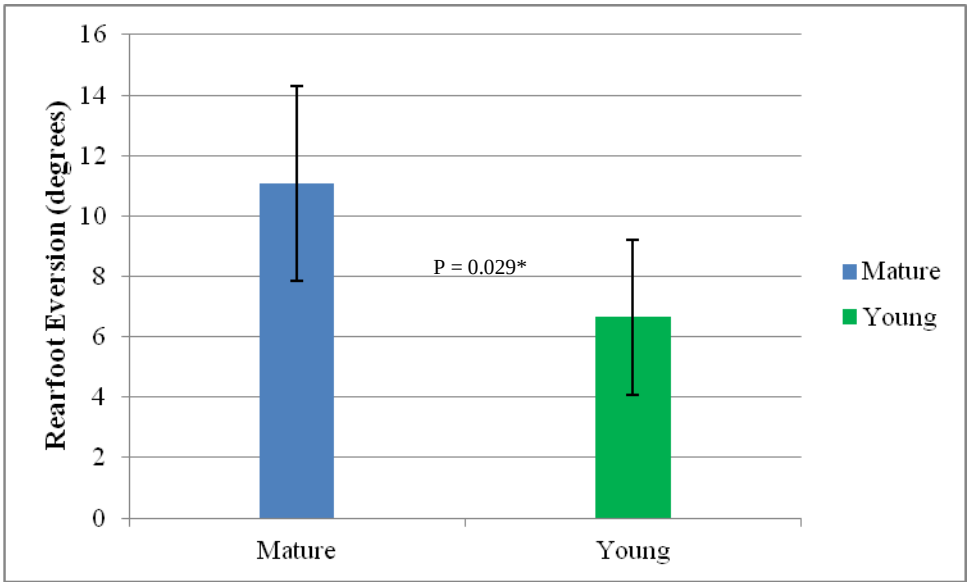


Figure 3.10. Bar graph illustrating significant difference in rearfoot eversion angle between the two groups.

3.2.2.4. Knee Internal Rotation Angle.

Figure 3.11 displays a bar graph highlighting the significant difference in knee internal rotation angle between the two groups. An angle time graph for knee internal rotation throughout the ground contact phase of gait is displayed in Figure 3.12. This displays a sample graph illustrating angle-time curves taken from the fifth trial of two individual subjects; one young and one mature. Calculation of group means showed a significant difference in the peak knee internal rotation angle between the two groups ($p<0.05$, Table 3.5).

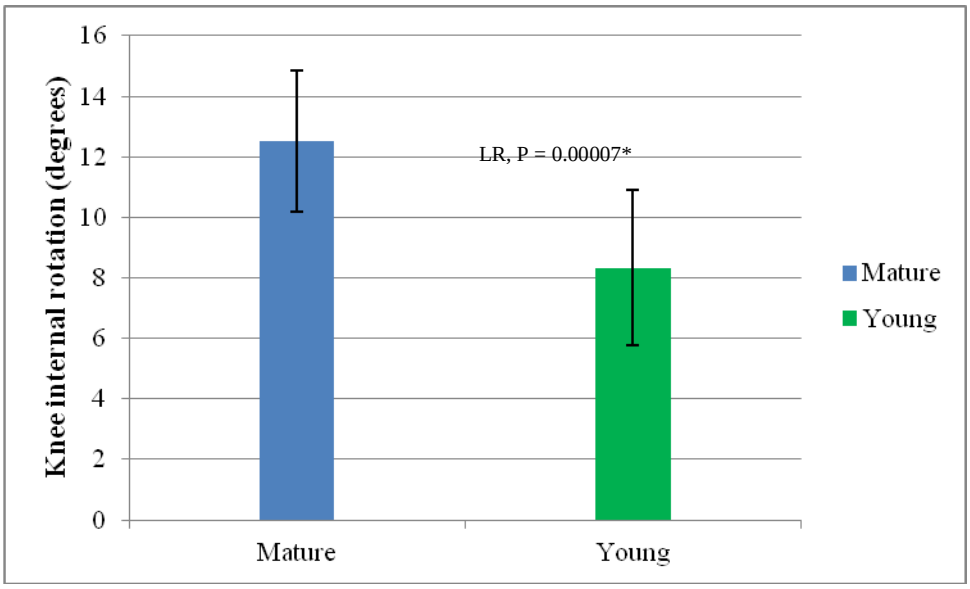


Figure 3.11. Bar graph illustrating significant difference in knee internal rotation angle between the two groups.

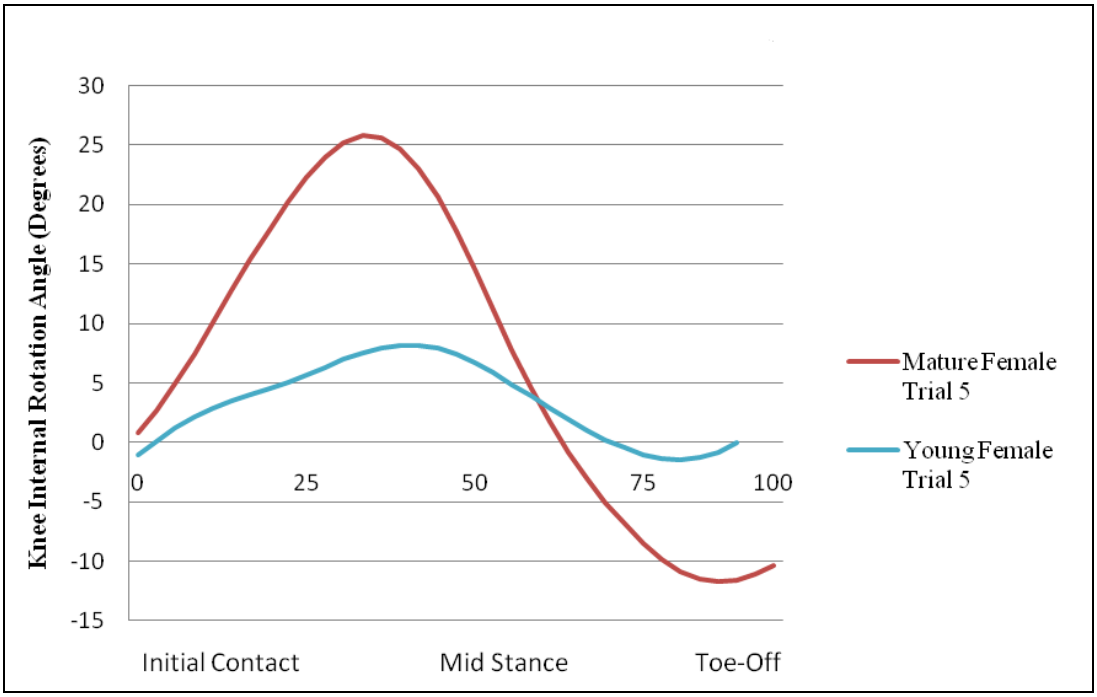


Figure 3:12. Graphical representation of knee internal rotation angle during ground contact produced by one sample subject from each group over one single running trial (Trial 5).

3.2.2.5. Peak ankle dorsiflexion angle.

Peak ankle dorsiflexion angle illustrated significant between group differences ($p<0.05$) (Table 3.5). Mature female runners showed an average peak value 4.6 degrees less than that produced by the younger runners. Figure 3.13 illustrates mean dorsiflexion angle trace for a single participant from each group. Figure 3.14 presents the dorsiflexion graphs produced by a mature female over the ten running trials. Data from two trials were omitted, as the graph suggested an adjustment in stride had occurred in order to contact the force plate. A bar chart of individual group means for each participant is presented in Figure 3.15, demonstrating an average trend for lower peak angles among the mature group.

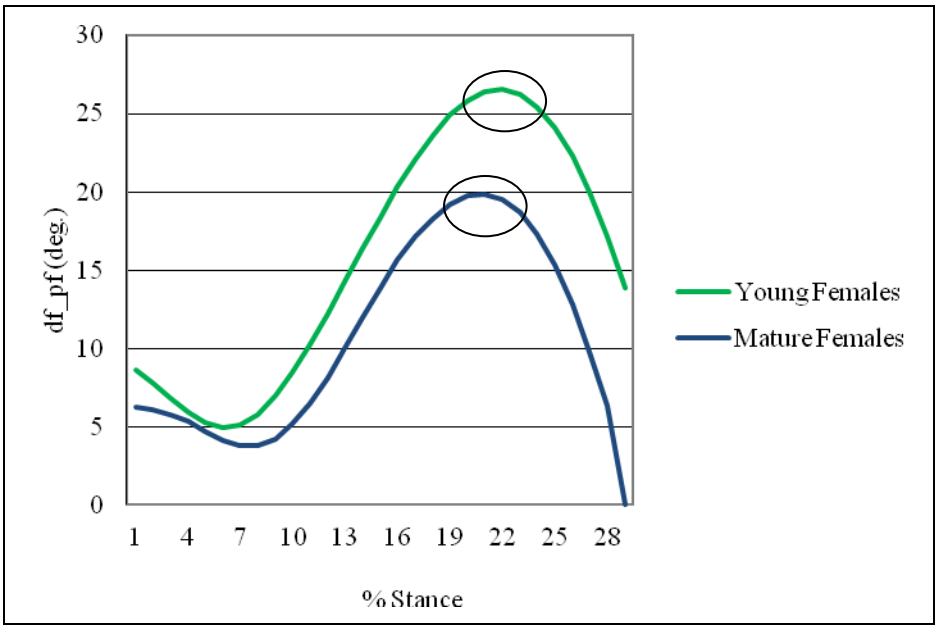


Figure 3.13. Ankle dorsiflexion angle throughout the stance phase of gait. Mean data for subject 5 (mature) and subject 9 (young).

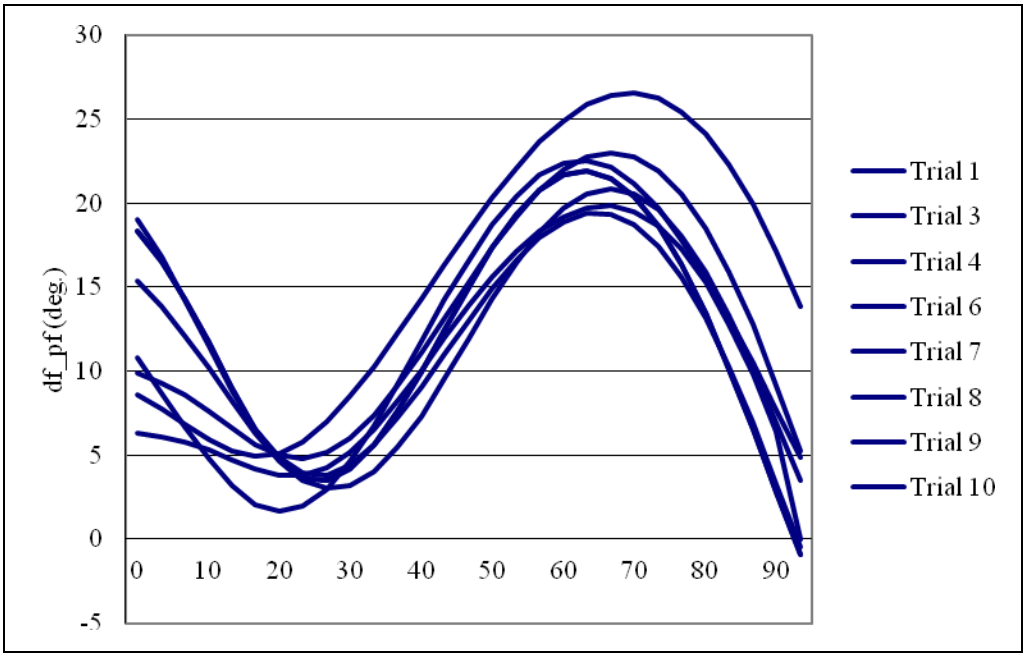


Figure 3.14. Dorsiflexion angle traces produced by participant 9 (mature).

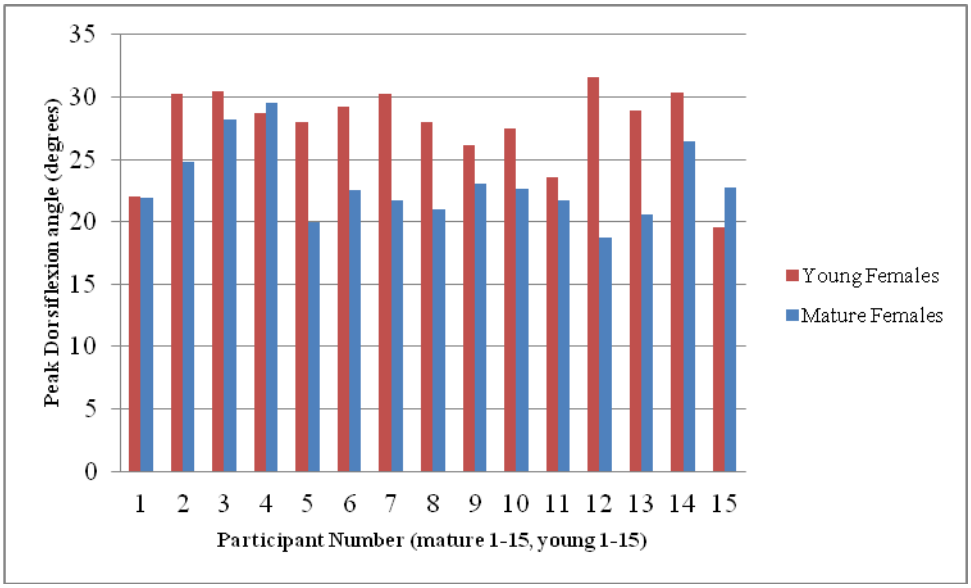


Figure 3.15. Bar graph illustrating individual mean data for peak ankle dorsiflexion angle throughout the stance phase of gait.

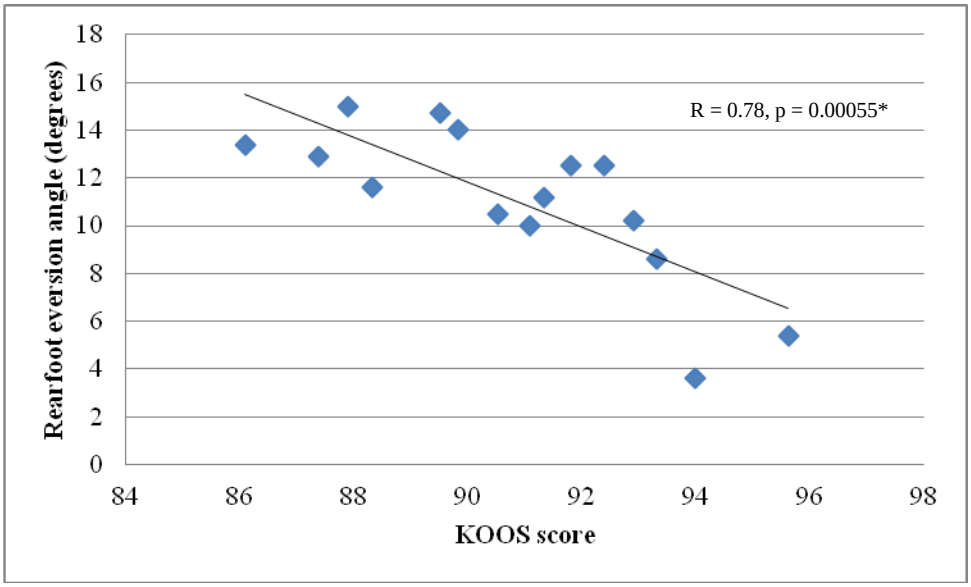
3.2.2.6. Correlation of Koos Scores with kinematic data.

Determination of a Pearson Product Moment Correlation Coefficient between KOOS score and kinematic variables are presented in Table 3.6. Where significant, correlations are presented in scatter diagrams (Figure 3.16a and b).

Table 3.6. Tabular presentation of results from a regression analysis between KOOS score and kinematic variables.

	Mean	Regression Analyses	
		R-Value	P-Value
Rearfoot eversion angle (degrees)	11.1 (±3.2)	0.78	**0.00055***
Knee internal rotation angle (degrees)	12.5 (±2.3)	0.50	**0.019***

(a)



(b)

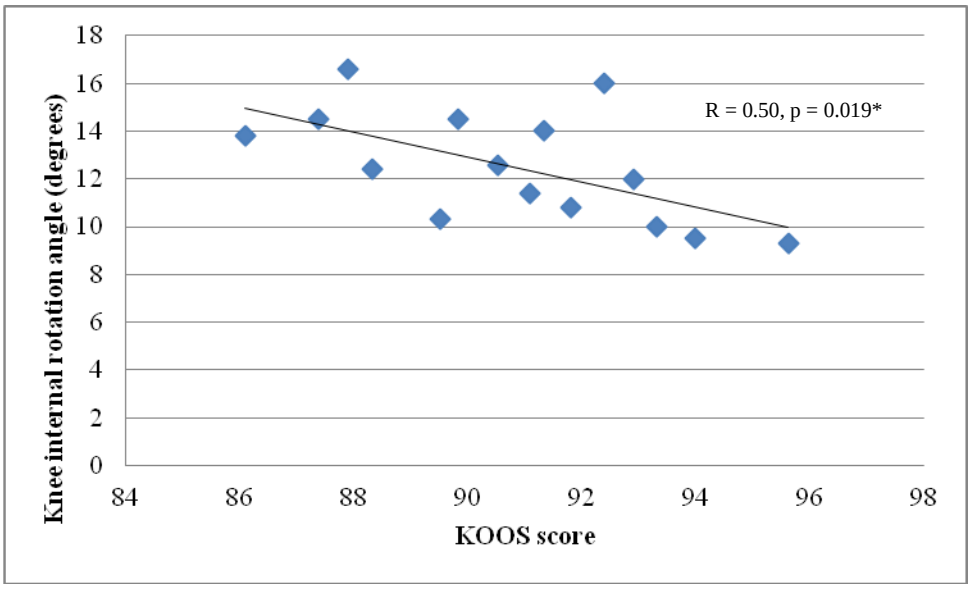


Figure 3.16 a and b. Correlation between KOOS score and rearfoot eversion angle (a), and knee internal rotation angle (b) among mature females.

3.2.2.7. Knee joint moments throughout the stance phase of running gait.

Knee joint moments were calculated in the frontal plane of motion during the ground contact phase of running gait. Group results are presented in Table 3.7. As shown, knee external adductor moments elicited an average between group difference of 0.58 Nm/kg; a value shown to be significant at the $p < 0.05$ alpha level.

Results from a regression analysis of the relationship between KOOS score and knee external adductor moment showed a significant positive correlation between the two variables ($r = 0.53$, $p = <0.05$). This relationship is highlighted through the use of a scatter diagram (Figure 3.17).

Table 3.7. Group means for the knee joint moments (Nm/kg) and occurrence times (s) calculated throughout the stance phase of running gait. All moments presented are normalised for body weight.

Group Data averaged from 15 subjects	**Frontal Plane Knee Moment**		
	Mean Peak (Nm/kg)	**Occurrence Times (s)**	
Mature subjects	1.75±0.25 (Adduction)	0.2 0.009	P = 0.00041*
Young subjects	1.31±0.34 (Adduction)	0.2 0.008	

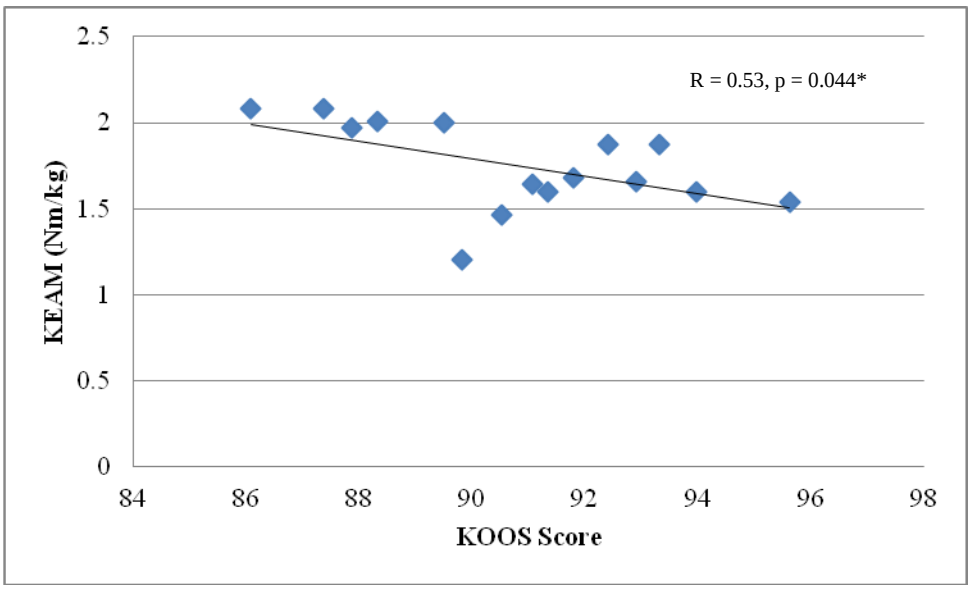


Figure 3.17. Correlation between KOOS score and knee external adductor moment.

3.4. Discussion.

3.4.1. Aims and hypothesis.

The main aim of the current study was to compare the running gait of young and mature females to investigate whether mature females demonstrate gait characteristics that may predispose them to the development of running related knee injuries and overuse conditions. With reference to related literature, it was anticipated that mature women (aged 40 years and above) would demonstrate significantly lower KOOS scores (indicative of more prominent knee symptoms) and different characteristics of running gait when directly compared to younger females. Specifically, it was hypothesised that mature women would demonstrate:

i. Greater rate of loading

ii. Greater knee internal rotation

iii. Greater rearfoot eversion angle

iv. Greater external knee adductor moments.

3.4.2. KOOS results.

Prior to initiation of gait testing, all females were assessed for the presence or potential of knee osteoarthritis via a questionnaire. All subjects both mature and young, were categorised as extremely low risk in all subscales; pain, symptoms, activities of daily living, sport and recreation, and quality of life, with an average score of 96.5 and 90.8 out of 100 for the young and mature respectively.

The knee osteoarthritis survey has been developed as an instrument to ascertain a person's opinion about the condition of their knee and associated problems. Although a range of other knee assessment questionnaires exist, Roos and colleagues showed the KOOS to be more sensitive and discriminative than the WOMAC assessment specifically the "sport and recreation" and the "quality of life" subscale (Roos, Roos, & Lohmander, 1999). This was based on an investigation of two groups; a group of patients diagnosed with osteoarthritis via radiography, and a group of controls. The KOOS scores showed significant differences between the two groups, demonstrating the efficiency of the survey in identifying osteoarthritis (Roos et al., 1999). Among the female runners, the results from the KOOS survey showed a significant difference between the two groups ($p<0.05$), with the mature group producing significantly lower scores (indicative of greater symptoms). Therefore, although both groups were categorised as free from signs of osteoarthritis, the results suggest a sensitivity of the survey to identify change in opinion of joint health with age.

A regression analysis was performed to ascertain the possibility of a relationship present between KOOS score and age among the mature females. Results illustrated an average negative correlation with age ($r = -0.6$, $p<0.05$) present on each subscale, the most prominent being the pain subscale; the greater the age, the lower the score. This therefore is consistent with Hascall & Kuetter, (2002) who suggested that osteoarthritis prevalence increases with age. Despite this, it is difficult to ascertain whether these results are expected, as currently no average score exists for different age brackets. In addition, a series of regression analyses were performed with the average KOOS score and the mean kinematic variables and moments produced by each participant (Figure 3.16 a and b; Figure 3.17). The observation that rearfoot eversion angle, knee internal rotation angle and knee external adductor moment were negatively correlated with KOOS scores ($r=-0.78$, $r = -0.50$, $r = -0.53$; respectively)

suggests that these variables may be linked with the eventual development of osteoarthritis or other conditions at the knee joint.

3.4.3. Kinetic variables: Peak impact force and loading rates.

Both kinetic and kinematic data were collected to provide a complete biomechanical description of lower extremity changes during gait, in attempt to characterise differences for two groups of different mean age. Following the scaling of all kinetic data to body weight, results from the female runners showed no difference in magnitude of the peak impact force between the two groups, suggesting no effect of age on this variable. Although little research has been carried out to investigate the effect of age group on ground reaction force variables, the results among the mature females appear contrary to one early study. In one investigation of walkers, Martin and colleagues demonstrated a change in age from a minimum of 27.5 to a maximum age of 73.5 years to elicit a significant decrease in vertical ground reaction force (Larish, Martin, & Mungiole, 1987). However it is noted that the results presented did not specify which phase of ground contact was measured, and the participants were not performing running trials. In contrast, a more recent study performed by Bus (2003) showed 16 older runners (55-65years) to exhibit significantly greater peak impact forces when compared to a group of younger runners (20-35 years). Here it was suggested that, although not measured, a difference in fat pad thickness under the calcaneus may have been a factor in these increased peak forces. More recently, it has been shown that peak vertical ground reaction force is lower among patients with symptomatic osteoarthritis of the knee joint (Childs, Sparto, Fitzgerald, Bizzani, & Irrgang, 2004). As such, the lack of change between the mature and young female runners suggests that the effect of age on this variable of gait is not certain.

The assumption that the impact peak of vertical ground reaction force can lead to overuse injuries are based on theoretical models of increased force and joint damage (Hreljac, 2004), however recent experimental studies appear to disagree. According to the results of a meta-analysis, Zadpoor and Nikooyan (2011) showed no association between the magnitude of ground reaction force and the development of stress fractures. Here it was suggested that the

ground reaction force cannot fully represent the loading of bones, and the impact force is, at most, only one risk factor for injury to the lower extremity. It is the internal force attenuation strategies (muscles, tendons, ligaments) that increase susceptibility to injury.

It has therefore been suggested that it is the rate at which the force occurs as opposed to the magnitude of the peak that has a greater influence on injury rates (Hawley, 2000). High rates of loading have been shown to be more damaging to the articular cartilage and subchondral bone than are forces of greater magnitude that are applied gradually (Kenneth & Brandt, 1997). As such, the vertical rate of loading during the impact phase of ground contact was assessed among the female runners. Results from the current study demonstrate an average loading rate of 54.9 ± 9.0 $BW.s^{-1}$ and 36.5 ± 8.9 $BW.s^{-1}$ for the mature and young female runners respectively, indicating the rate at which load is transmitted to the lower extremity (Aguinaldo & Mahar, 2003). With a significance observed at the $P<0.05$ level, the higher value observed for loading rate of ground reaction force for the mature females than for the younger, suggests shock absorbency is relatively poor for mature females, which could relate to poor function at the knee joint (Richards, 2008).

Loading rates can be harmful to the joint, if magnitudes exceed the physiologic tolerance level for an individual. In an early mechanical test of bones, it was discovered that the fatigue stress of bone (indicating the level of fatigue the bone can handle prior to damage) was significantly lower when load was applied at a higher strain rate (Schaffler, Radin & Burr, 1989). As such, it could be suggested that high loading rates could increase the risk of running related injuries (stress fractures) to the bones of the lower limb. According to Syed and Davis (2000), high loading rates can also lead to articular cartilage damage, suggesting that the mature females are at higher risk of developing conditions such as osteoarthritis. This variable has been associated with accelerated progression of existing osteoarthritis, and onset of osteoarthritis at previously asymptomatic joints (Mundermann, Dyrby and Andriacchi, 2005). It has been shown that rapid acceleration of load does not allow sufficient time for the periarticular muscles, the major shock absorbers protecting the joint, to absorb the load (Scott, Menz & Newcombe, 2007) suggesting that the joint is subsequently subjected to a higher load.

The greater loading rate of ground reaction force observed for the mature females compared with younger female runners in the current study therefore suggests that investigation of an intervention to reduce loading rates in this population would be beneficial. If this could be

achieved, it may reduce the incidence or development of running related injuries such as stress fractures and overuse conditions including osteoarthritis. The most common approach to reducing this variable has been the introduction of a cushioned heel in footwear, either through the selection of shoes with specific heel cushioning, or through the addition of a cushioning insole within the shoe (Mundermann, Stefanyshyn & Nigg, 2001). Whilst conflicting evidence exists regarding the success of this approach in reducing loading rate (Withnall, Eastlaugh, & Freemantly, 2006), it is suggested that the potential of this intervention in reducing this variable in mature female runners should be investigated further.

3.4.4. Kinematic variables and risk of injuries and overuse conditions.

3.4.4.1. Standing Q angle data.

The observation of no significant difference in Q angle between the mature females and the younger group indicates that relative anatomic angle does not alter with age among this population of females. These results support those presented by Hsu and colleagues, who, despite illustrating a significant effect for gender, showed age to have little effect on factors relating to axial alignment of the lower extremity (Hsu, Himeno, Coventry, & Chao, 1990). In addition, it has previously been speculated that high rates of participation in activities that involve quadriceps contraction, causes the Q angle to straighten or decrease in magnitude (Hahn, & Foldspang, 1997). As such, it could be supposed that the similar level of running undertaken by the females in each group may be an additional reason for the similar quadriceps angles produced by the two age groups.

The Q angle is a measurement of patellofemoral joint biomechanics, commonly used in musculoskeletal medicine, as an indication of uneven load at the knee joint, and changes in lateral compartment pressure (Herrington & Nester, 2004). It has been demonstrated in the literature that females present a greater Q angle when compared to males, however little research has investigated the effect of age. The Q angle values calculated among the mature and young females are low in comparison to those presented in the literature. In an evaluation of lower extremity running injuries, Hamill and colleagues categorised high and low Q angles

as those calculated to be above and below 15° respectively (Hamill et al., 1999). It could be considered that differences in calculation methods may affect the Q angle produced, as within this investigation, the angle was measured from the greater trochanter, not the anterior superior iliac spine as suggested in the literature (Kirtley, 2006). However as a standardised method was used to calculate this angle for all females in this study, the particular method used has little effect on group comparisons, and the lack of difference between the two groups suggests little effect of age among the females in the current study.

3.4.4.2. Kinematics during running.

Examination of the kinematic data collected in the present study indicated a significant difference in biomechanical movement patterns of young and mature females. Within this investigation, mature females demonstrated an average peak rearfoot eversion angle of 5.6^{o} greater than the young females ($p<0.05$), which is consistent with the literature suggesting that greater rearfoot eversion angles are commonly associated with increasing age (Scott, Menz and Newcombe, 2007). As shown in Figure 3.9, the rearfoot progresses from an inverted position at heel strike, to maximum eversion during early to mid stance. As all females were categorised as heel strikers, this movement occurred as the forefoot contacted the ground during early midstance, the body weight was positioned over the foot, and the subtalar joint pronated to absorb impact forces and allow the foot to adapt to the terrain (Maffulli, 2001).

One possible explanation for these high eversion angles could be the increased laxity of the ligaments within the foot and ankle that has been shown to occur with age. Although the association between arch height and subtalar pronation is not equivocal (Nigg et al., 1993), according to Nigg and colleagues, laxity of the plantar ligaments supporting the arch of the foot was higher among a group of mature compared to younger females (Nigg, Fisher, & Ronsky, 1994).

It is suggested that the large magnitudes of rearfoot eversion produced by the mature females could predispose this group to injury and debilitating conditions at the knee joint. High eversion angles and accompanied loading under the medial forefoot have been associated

with injuries such as patellofemoral disorders and iliotibial band friction syndrome, and the development of exercise related lower leg pain (Willems, DeClercq, Delbaere, Vanderstraeten, DeCock & Witvrouw, 2006; Clanton, 1992). Similarly, a strong association has been shown between flat-footed individuals and the incidence of knee joint osteoarthritis (Reilly, Barker Shamley & Sandall, 2006); although no direct causative mechanism between foot structure and osteoarthritis development was concluded. It is suggested that flat-footedness may increase subtalar joint pronation due to increased laxity of the ligaments in the ankle, subtalar joint and medial longitudinal arch that would enable the latter structure to collapse. These results contribute to the justification for exploring the role of footwear in the development of lower extremity conditions. Although not definitive in the literature, footwear techniques have been proven to alter rearfoot kinematics during running, and will therefore be explored in the following study among mature female runners.

According to Bus (2003), age has been shown to have little effect on rearfoot eversion, or range of motion, among male runners aged 55to 65 years (Bus, 2003). Based on the results from the current study among female runners, it could be considered that age related changes in running gait occur during the third and fourth decades. As the standard deviations among the females in this investigation were low, it is suggested that the results produced do highlight a difference in rearfoot mechanics for mature female runners compared with younger females.

An increased magnitude of subtalar joint pronation has also been associated with increased knee internal rotation and suggested to increase lateral knee loading (Olson, 2007). Hintermann and Nigg (1998) suggested it is not only the amount of rearfoot eversion, but the transfer mechanism of eversion into tibial internal rotation that is crucial to knee joint stress, and leads to lower extremity overloading. Felson *et al*., (2000) support this, suggesting greater eversion of the subtalar joint causes the knee to assume a more internally rotated position. This conflicts with external rotation during knee extension as the body weight passes over the foot, increasing pressure on the soft tissue of the knee joint.

The present study supports these theories with significantly higher knee internal rotation angles also observed for the mature females ($p<0.05$). As illustrated in Figure 3.12, throughout the stance phase of one single running step, for both groups the knee moves from a slightly externally rotated position, to peak internal rotation during the first half of stance. Following this, the knee then returns to an externally rotated position, which coincides with

the peak knee external adductor moment occurring during mid to late stance. Peak internal rotation angles were identified as 12.5 ± 2.3^{o} and 8.3 ± 2.57^{o} for the mature and young females respectively. According to Beckman (1980), excessive internal rotation of the tibia and knee interferes with the normal knee-muscle force vectors, and shifts the patella medially, causing increased compressive forces to act on the knee joint. Additionally, Hunter, Dolan and Davis (1995) suggested that large values of internal knee rotation could increase the force between the lateral aspect of the patella and the lateral femoral condyle, increasing lateral knee pressure. Within their research, Hunter *et al*., (1995) further suggested a link between this internal rotation and increased Q angles, which cause displacement of the resultant force acting at the knee joint, increasing the irritation to the articular cartilage of the knee. Although the standing Q angles were not different between the two groups, the knee internal rotation during the stance phase of gait likely to have produced an increase in Q angle. Therefore, in light of the present results, alongside a review of relevant literature, both the greater ankle eversion angle, alongside the increased knee internal rotation displayed by the mature female runners are deemed potential factors predisposing to conditions at the knee joint.

Results from this investigation showed mature females to run with significantly smaller angles of peak dorsiflexion during the stance phase of gait ($p<0.05$). Among heel strikers, dorsiflexion occurs during both the heel contact and mid stance phase of running gait. As illustrated in Figure 3.13, ankle dorsiflexion at heel contact was followed by plantarflexion during early stance. Peak ankle dorsiflexion occurred during mid to late stance, followed by the second peak of ankle plantarflexion in preparation for toe off. It is noted here that variability within each participant's data was seen in the dorsiflexion at initial contact. As shown in Figure 3.14, a variation in initial contact angle of up to 10° could be seen between trials, eliciting large standard deviations in the means. As such, this variable at initial contact was not assessed between groups.

During the stance phase of running gait, dorsiflexion combines with subtalar joint pronation and internal rotation of the tibia to act as a shock absorbing mechanism (Novacheck, 1998). According to Brukner & Khan (2006), dorsiflexion is required during gait to enable the tibia to rotate over the foot during the stance phase, without early heel lift or excessive subtalar pronation occurring to compensate. Angles of peak dorsiflexion below 15 degrees are occasionally considered a consequence of increased tightness of the gastrocnemius, and may

cause the subtalar joint to excessively pronate in order to utilise the dorsiflexion component of pronation (Brukner & Khan, 2006). Although the mean angle produced by the mature females was 23.04 ± 3.01^{o}, as an average, these values are considered on the lower side of ideal. Comparative studies in the literature have illustrated average dorsiflexion values among uninjured runners to be around 25° to 30° (Novacheck, 1998; Hamill & Knutzen, 2003), a value higher than that produced during both walking and sprinting. As the average value for the younger group was 27.6^{o}, the lower values for the mature group could suggest decreased shock absorbing ability among this latter cohort.

The effect of age on biomechanics during gait has previously shown mixed results, with general theories suggesting that both posture and balance are the main variables that change with increasing age (Kim, Lockhart and Yoon, 2005). Furthermore, it has commonly been suggested that mature participants tend to adopt a slower gait, and decrease stride length (Kim et al., 2005). This notion was supported in a sample study by Bus who showed a group of more mature runners to automatically adopt a slower self selected running speed compared to a group of younger participants (Bus, 2003). Among the female runners in the current study, the speed of gait was controlled, reducing the possible effect on gait. As data were based on one single running step, stride length was not measured. It is suggested that the gait differences displayed by the mature and young females were unaffected by the speed and stride changes that occur with age.

3.4.5. Knee joint moments.

Previous research has suggested that the peak knee external adductor moment is an indirect measure of dynamic knee joint loading during gait; therefore this was assessed during running and compared between mature and young females (Andriacchi, 1994). As displayed, the mature females demonstrated an average moment 33.1% greater than the young; a possible cause of greater malalignment among this group (Table 3.7). Figure 3.18 illustrates a diagram of the external adductor moment acting at the knee joint; influenced predominantly by the calculation of the moment arm from the knee joint centre to the line of ground reaction force, and the magnitude of that force (Riskowski, Dufour, & Hannan, 2011).

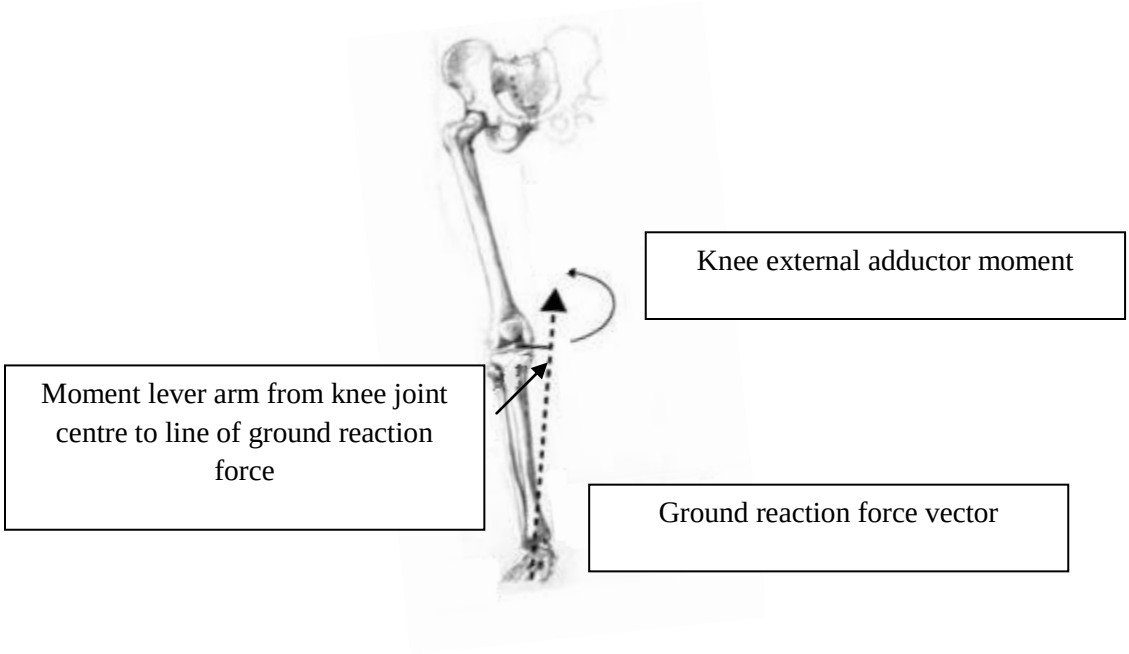


Figure 3.18. Knee external adductor moment acting on the right knee joint. Source: Adapted from Riskowski, Dufour & Hannan, Current Opinion in Rheumatology, (2011).

The knee external adductor moment (Figure 3.18) is generated by the combination of the ground reaction force passing medially to the centre of the knee joint, and the perpendicular distance from the line of this force to the knee joint centre (Andriacchi, 1994). This occurs when the knees are in a varus position, and the moment acts to adduct the knee joint increasing the load on the medial compartment (Russell and Hamill., 2011). As such, the knee external adductor moment can be used as a valid and reliable representative of the medial to lateral load distribution at the knee joint (Birmingham et al., 2007). The first to propose this theory were Schipplein and Andriacchi, who predicted medial compartment loads with a statically determinate muscle model, and suggested the knee external adductor moment was the primary predictor of dynamic medial compartment load (Schipplein & Andriacchi., 1991). Following this, Noyes and colleagues illustrated a statistically significant correlation between medial compartment load and the knee external adductor moment among ACL deficient patients, and Hurwitz and colleagues later significantly associated the knee external adductor moment with the distribution of bone between the medial and lateral sides of the proximal tibia (Noyes, Schipplein, Andriacchi, Saddemi & Weise, 1992; Hurwitz et al., 1998). Finally, Zhao and colleagues demonstrated a significant positive correlation between

knee external adductor moment and medial compartment loads measured internally (Zhao, Banks, Mitchell, D'Lima, Cowell., & Fregly, 2007).

When large moments present at the knee joint, the articulation between the medial aspect of the tibia and the inferior medial femoral condyle experiences increased loads (Hunt *et al*., 2006). As such high knee external adductor moments have been consistently associated with features of medial compartment knee osteoarthritis, including joint space width, disease severity and progression (Miyazaki, Wafa, Kawahara, Sato, Baba & Shimada, 2002). For example higher baseline moments were observed among patients who went on to develop radiographic progression of osteoarthritis compared to those that did not (Hurwitz, Summer, Andriacchi, & Sugar, 1998). Furthermore, Bolt and Morag (2007) have even used peak adductor moments as a direct reflection of internal loads at the knee joint when assessing the effects of orthotics, supporting the association of moments and overuse injury potential.

Age, previous joint injury, and malalignment are well-known risk factors for the progression of osteoarthritis at the knee joint. The knee external adductor moment has been associated with osteoarthritis development both among severe and early stage patients. According to Erhart and colleagues, the knee external adductor moment has been associated with severity, rate of progression and treatment outcomes of medial compartment knee osteoarthritis (Erhart, Mundermann, Elspas, Giori & Andriacchi, 2008), and Sharma et al., (2001) significantly associated knee external adductor moments with disease severity among symptomatic OA patients. Other longitudinal investigations have looked at the effect of knee external adductor moments over time, and one study has showed the value of knee moments at baseline to predict disease progression in the medial compartment of the knee joint over a six year time frame (Miyazaki et al., 2002). Consequently, due to this association between knee adduction moments and osteoarthritis, this biomechanical variable has become a potential method to assess treatment outcomes. Many surgical outcomes and biomechanical interventions are aimed at reducing the knee external adductor moment, to represent reduced loading of the medial knee (Erhart et al., 2008).

Although the external adductor moment has commonly been investigated among osteoarthritic patients, an association between the condition of the knee and the magnitude of the knee external adductor moment has been shown among healthy mature females. In one investigation, Davies-Tuck and colleagues demonstrated a significant correlation between meniscal tears and knee external adductor moment, showing a link between this variable and

early signs of joint degradation (Davies-Tuck, Wluka, Teichtahl, Martel-Pelletier et al., 2007). Among the mature female runners of the current study, the results from the knee osteoarthritis observation survey showed a significant correlation between KOOS score and knee external adductor moment (Figure 3.17). Although all females were categorised as free from symptomatic osteoarthritis, this significant correlation suggests increased investigation is required into the association between asymptomatic medial knee osteoarthritis and the knee external adductor moment, perhaps using an alternative method for the quantification of osteoarthritis presence.

A strong association between knee external adductor moments and medial compartment knee joint osteoarthritis has been verified in the literature. As a result of this, the majority of research has focussed on this biomechanical variable among more mature, osteoarthritis prone, individuals. Therefore the direct association between knee external adductor moments and age has not been extensively investigated. Rudolph and colleagues showed a change in medial-lateral joint laxity among a group of elderly participants during walking, and increased knee adduction, although the age range was above that of the mature females in this study (Rudolph, Schmitt, & Lewek, 2007). It is suggested that the main association between knee external adductor moment and age is due to the theory that osteoarthritis risk increases with age. According to Loeser (2010) the concept that aging contributes to but does not directly cause osteoarthritis is consistent with the multifactorial nature of the condition. It is suggested that the high external adductor moments produced by the mature female runners when compared to the younger runners in the current study could be a main cause of the high risk of overuse conditions among the former group, particularly the incidence of medial compartment knee osteoarthritis. The increased medial load indicated by the high knee external adductor moment is in addition to the high lateral knee load suggested by greater rearfoot eversion and knee internal rotation for this cohort. However, the relationship between rearfoot movement and localized knee joint loading requires further investigation.

3.4.6. Limitations.

A limitation in this investigation was the difference in age range of the two groups, with a range of 20 years and 6 years for the mature and young group respectively. This occurred due to difficulty in recruiting mature females, and a need to maintain similar group sizes. Additionally, the runway used in this study was only 10 metres long, which may have limited each participant's ability to attain a natural, consistent running style. However, all attempts were made to overcome this potential limitation, and familiarisation trials occurred to enable participants to run with a natural gait. Furthermore, participants were encouraged to repeat trials did not feel reflective of their natural style. Force plate targeting was an additional potential limitation; however participants were permitted to mark the floor once a comfortable start position had been identified. Each female would track the pace from the start position to ensure a natural running pace could be achieved, and force plate contact could occur relatively effortlessly. Furthermore, during data analysis, horizontal velocity values were assessed to ensure a constant pace was adopted. Finally, the footwear was not-standardised. Participants wore their own trainers, in an attempt to reflect their natural running style. As comparisons were made been ankle and foot biomechanics between the two groups, it is considered that the different footwear worn by each participant may have altered the gait. This in combination with the abundance of previous research that has investigated the influence of footwear on running biomechanics, gave way to a possibility of including footwear as a variable in future research within this group.

3.5. Conclusion.

This investigation was undertaken to compare the running gait of young and mature females to investigate whether mature females demonstrate gait characteristics that may predispose them to the development of running related knee injuries and overuse conditions. The pathogenesis of both overuse injuries and knee joint osteoarthritis is complex and involves many correlated factors that can be measured with gait analysis (Astephen & Deluzio, 2003). A review of relevant literature suggested that certain gait parameters are closely related to potential injury and osteoarthritis development, including rate of loading of impact force, knee internal rotation and subtalar joint pronation (Norris, 2004). Furthermore, high external knee adductor moments have consistently been demonstrated among osteoarthritis patients, and been described as a significant risk factor for disease instigation (Kaufmann, Hughes, Morrey, Morrey, & An., 2001; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio., 2007).

The results from this investigation support the proposed hypotheses; with higher values of loading rate, internal rotation of the knee, ankle eversion and knee external adductor moments observed for the mature females. In addition, lower rates of ankle dorsiflexion were shown among mature females. This study lies among several other research papers in the categories of overuse injuries and knee osteoarthritis. The results of the current study, combined with the weight of the research literature, lend support to the initial notion that the running gait of mature females exhibits certain characteristics that may predispose to lower limb and knee injuries, and medial compartment knee osteoarthritis. Although results from the KOOS survey provide little evidence of osteoarthritis present among the mature females, it is possible that osteoarthritis or other conditions may develop over time if these runners continue to exhibit such gait patterns.

3.6. Progression and further research.

3.6.1. Summary of background.

The first study within this research thesis produced the following summarised conclusions. When comparing the biomechanical components of running gait between a group of young (18-24 years) and a group of mature (40-60 years) females, significant differences were observed. Kinetic analysis of the ground reaction force variables displayed a significant difference in the loading rate data. The rate of loading during the impact phase of one running stride was significantly greater for the mature women compared to the younger. The group means for the peak knee internal rotation angle also showed significantly greater angle magnitude for the mature group, as did the group means for the peak rearfoot eversion angle exhibited during the ground contact phase. Group means for peak ankle dorsiflexion during mid stance were shown to be significantly lower among the mature group. Finally, the frontal plane knee moments (peak external adductor moment) produced during running were on average 33.1 % higher among the mature females compared to the younger. A review of relevant literature has suggested that each of these gait parameters have previously been related to the initiation of joint deterioration and osteoarthritis development (Norris, 2004; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio., 2007). Therefore, in light of the conclusions drawn, it appears clear that further investigation is required within this specific population.

Chapter 4.

"The Influence of Motion Control Shoes on the Running Gait of Mature and Young Females."

4.1. Summary of Previous Study Findings and Conclusions.

As the age of female runners increases, certain biomechanical features of running gait are subject to change. When compared to a group of younger runners, mature females have been shown to run with increased rearfoot eversion, knee internal rotation, high loading rates, and large knee external adductor moments (Lilley, Dixon & Stiles, 2011). A discussion of results combined with those presented in the relevant literature suggested that each component of gait could predispose this group to running injuries. As a commonly accessible form of altering alignment during running, the effect of a motion control shoe on each component of gait could provide insight into possible injury prevention techniques for this specific group.

4.2. Introduction and Review of Relevant Literature.

4.2.1. Incidence of running injuries among mature females.

Both gender and age have put mature female runners at a high risk of overuse injuries to the lower limb (Taunton et al., 2002), and the development of knee joint osteoarthritis (Hunt et al., 2006). Furthermore, it has also been noted that as the age of this population of runners increases, the running gait patterns are subject to change (Ferber et al., 2003). The relationship between advancing age and biomechanical factors influencing risk of overuse injuries and conditions among females was therefore investigated in the previous chapter (Lilley, et al., 2011). This study directly compared the running gait of mature (mean 48.2 ± 3.6 years) and young (mean 22.5 ± 4.8 years) females. Results from this initial study

illustrated mature females to demonstrate a significantly different running pattern compared to the younger females ($p<0.05$). Specifically, the mature group were found to exhibit significantly greater rate of loading, peak knee internal rotation angle, peak rearfoot eversion angle, decreased ankle dorsiflexion, and increased peak knee external adductor moment, during the stance phase of running ($p<0.05$). Discussion and interpretation of these results with reference to related relevant literature suggested a link between each variable and the potential for injury occurrence and the development of knee joint osteoarthritis. Specifically, the high knee external adductor moment produced by the mature female runners was associated with the potential for injury occurrence and increased medial knee joint osteoarthritis risk (Hunt et al., 2006).

4.2.2. Excessive rearfoot eversion, knee internal rotation and loading rates during running.

As the foot is represented as a rigid structure, rearfoot eversion is commonly accepted as the dominant and therefore representative component of subtalar joint pronation. Mature female runners have been shown to run with high magnitudes of loading rate, rearfoot eversion angle, and knee internal rotation when compared to a group of younger female runners. As discussed in the previous chapter, excessive rearfoot eversion and knee internal rotation have commonly been linked with increased risk and development of running related injuries. At the distal end of the tibia, subtalar joint pronation allows for the absorption of impact forces, and without this movement, the forces would have to be absorbed abruptly causing a high level of stress on the supportive structures (Bates et al., 1978). However, in excessive magnitudes, rearfoot eversion causes increased rotation of the tibia and knee due to the coupling between these joints. This results in overloading of the knee joint and increased injury risk (Hintermann and Nigg, 1998).

Rate of loading of ground reaction force during running describes the velocity with which the ground reaction force vector is attenuated. A detailed description of ground reaction force and loading rate calculation is described in Chapter 2 (2.5.1, Equation 2.4). According to Stergiou and colleagues, the repetitive encounter between the ground and the foot during running results in impact forces that must be absorbed by the supportive structures of the lower limb

(Stergiou et al., 1999). During heel strike running gait, these forces can be subdivided into the impact forces as the segments decelerate at ground contact, and the active forces during push off phase (Nigg, Bahlsen, Luethi, & Stokes, 1987). The peak of the impact force is defined as that which occurs within the first 50 milliseconds after the heel has contacted the ground (Nigg et al., 1987). During this impact phase, the momentum from the decelerating limb alters at ground contact, resulting in the transient force transmitted up the skeleton (Aguinaldo & Mahar, 2003). Although early research commonly considered that high impact forces during running were associated with pain, injuries and damage to the articular cartilage (Clement et al., 1981), more recently it has been suggested that it is the rate of loading of this impact force that is more strongly associated with injury risk (Hawley, 2000). Similarly, Nigg later suggested that combined theoretical, experimental and epidemiological evidence on ground reaction impact force did not lend support to an association between this variable of gait and chronic and/or acute running related injuries (Nigg, 2001). Despite this, although not uniformly conclusive, the loading rate of impact force during running and walking has previously been linked with injury risk and development. According to Nigg and colleagues, high loading rates are linked with periodic stretching of the soft tissues in the joints, which in turn is connected with pain and injuries in the lower leg (Nigg, Bahlsen, Luethi & Stokes, 1987). In support of this, Hennig and Lafortune associated increased loading rates with higher levels of shock transmitted along the lower extremity, and Grimston and colleagues linked loading rates to increased risk of bony injuries of the lower limb (Hennig & Lafortune., 1991; Grimston, Ensberg, Kloiber, & Hanley, 1991).

In light of these theories presented in the literature, it has become clear that the running style associated with mature females could be a contributing factor to the vast level of injuries and overuse conditions among this particular sub population. However, prior to an investigation of possible injury reduction techniques, both quadriceps strength and knee joint stiffness were considered in their contribution to biomechanical gait related changes with age. Strength of the quadriceps muscle is assessed using isokinetic technologies, and is an important factor in the control of knee joint motion during running (Norris, 2004). Knee joint stiffness implies a reduction in knee range of motion in the sagittal plane, and has been linked with the development of knee joint osteoarthritis (Butler, Crowell, & Davis., 2003).

4.2.3. Knee flexion and injury risk.

It is clear that the flexion extension movement of the knee is cyclic, precisely controlled by the predominant muscles in the upper leg. However gait abnormalities can involve the failure of one or both of these actions to complete their required motion (Whittle, 2003). When investigating the differences in maximum flexion angle between injured and non injured groups, previous research has indicated varied results. Baliunas and colleagues indicated greater knee flexion during walking in subjects presenting medial compartment knee osteoarthritis compared to controls (Baliunas, Hurwitz, Ryals, Karrar, Case, Block and Andriacchi, 2002). Similarly, a later study by Heiden and colleagues showed a group of physician diagnosed osteoarthritis patients to exhibit significantly greater flexion angles at heel strike, early and late stance when compared to a group of asymptomatic controls (Heiden, Lloyd, & Ackland, 2009). However this is not a consistent finding among osteoarthritis patients, and Mudermann and colleagues (2005) showed greater knee flexion among a group of osteoarthritis patients when compared to controls (Mundermann, Dyrby, & Andriacchi, 2005). Zeni and Higginson (2009) showed no significant difference in sagittal plane knee joint motion between patients with moderate osteoarthritis and controls, whereas a group of patients with severe osteoarthritis showed reduced knee range of motion.

Therefore it seems that in comparison to other gait parameters such as rearfoot eversion, the knee flexion/extension movements are less definitive in their role as injury mechanisms. It could be suggested that variation in findings when investigating diagnosed osteoarthritis patients could be due to severity of the condition, length of time since diagnosis, and specific location of the pathology. This is further highlighted in a direct comparison between the study performed by Messier and colleagues who investigated ten osteoarthritis patients with varying severities in all three compartments of the knee joint, and Schmitt and Rudolph who investigated twenty eight patients with osteoarthritis confined to the medial compartment (Messier, DeVita, Cowan, Seay, Young & Marsh, 2005; Schmitt & Rudolph, 2007). Results from the former study illustrated no difference in knee kinematics and kinetics between osteoarthritis patients and controls, whereas results from the latter study showed significantly reduced range of motion among the osteoarthritis patients compared to the controls.

Some research has identified a link between the magnitude of peak flexion and extension angles during gait and certain injuries and loading. Boden, Dean, Feagin and Garrett (2000)

investigated knee injuries, specifically injuries to the ligaments, among a group of athletes. Their findings suggested that the majority of injuries occur during knee extension phases of gait, as the quadriceps contract concentrically and power is generated (Novacheck, 1998). Additionally, results indicated greater laxity in the hamstring flexibility parameters measured among the injured, indicative of reduced flexion control and subsequent abnormal knee motion during running. This notion that reduced flexion control may be linked to injury was further supported by Noyes, Dunworth, Andriacchi, Andrews and Hewett (1996), who demonstrated an increase in knee flexion and related decrease in extension moment during the stance and terminal stance phases respectively, resulting in a significant decrease in the calculated medial tibiofemoral loads among symptomatic patients with ligament complex injury. Similarly, it has been suggested that increasing knee flexion angle may reduce ligament loading, thereby reducing the risk of injuries to the knee joint (Cochrane, Lloyd, Besier & Ackland, 2003). Cochrane et al., (2003) later supported this hypothesis, showing a strength and balance training programme to significantly increase knee flexion angles and decrease internal knee loading (Cochrane, Lloyd, Besier, Ackland & Elliot, 2003). This theory has been investigated during landing movements, where reductions in knee flexion angle are more common among females, and can increase risk of injury to the knee joint (Lephart, Ferris, Riemann, Myers, & Fu, 2002). Although not investigated during running, these authors suggest that weaker thigh muscles and reduced flexion control may cause an increased stiffness of the knee and lower leg during landing, and increase risk of injury. This was supported by Malinzak and colleagues, who showed females to exhibit reduced knee flexion angles when compared to men during running, side-cutting and cross-cutting activities suggesting different motor control patterns that may increase female risk of injuries to the knee joint (Malinzak, Colby, Kirkendall, & Yu, 2001). Although previous literature findings suggest an ambiguous link between knee flexion angle and injury, the literature above illustrates a notion that reduced flexion during gait can increase the internal loading at the knee joint; a possible mechanism of injury.

4.2.4. Knee abduction angle and injury risk.

As described in section 2.4.5.2, although the medial and lateral collateral ligaments are in place to increase the stability of the knee joint, certain deformities and malalignments can occur during standing and gait. This can therefore alter the inherent load attenuation strategies present in the lower extremity. During the flexion and extension movements of the knee throughout gait, it has been suggested that the patella should ideally travel in line with the long axis of the femur (Norris, 2004). However, in a malaligned knee, the tracking of the patella could be slightly distorted, subjecting the patella to minor dislocations during gait (Norris, 2004). Furthermore, during running high loads are applied to the lower extremity, and alignment is a key determinant of load distribution at the knee joint (Sharma, Song, Felson, Cahue, Shamiyeh, and Dunlop, 2001). These loads are transmitted through the mechanical axis of the lower extremity; a line progressing from the centre of the head of the femur to the mid superior talus (Johnson, Leitl, & Waugh, 1980). In the normal knee, this line passes through or very near to the centre of the knee joint (Maquet, 1984). This is highlighted in the diagram (Figure 4.1).

In theory a shift from the collinear alignment of the hip knee and ankle affects the load distribution at the knee; deviations to the position of the line can imply excessive load bearing on one compartment. In a valgus knee, the contact of the bone surfaces are uneven with reduced contact on the medial side. Therefore as the forces are transmitted through the knee, higher loads are applied to the lateral knee, increasing the wear on this area of the joint, and increasing the susceptibility to related conditions such as lateral compartment knee joint osteoarthritis (Johnson et al., 1980).

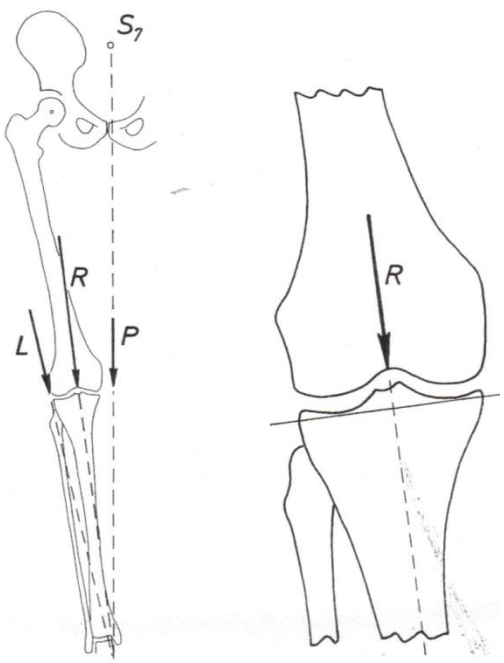


Figure 4.1. Load (R) acting through the centre of the knee joint in a neutrally aligned knee. Source: Maquet, (1984).

Early research by Maquet (1976) illustrated the significance of varus and valgus deformities in the coronal plane to the development of osteoarthritis of the medial and lateral compartments of the knee respectively. This notion was later supported by Sharma, Song and Felson (2001), who performed a longitudinal study to assess the varus and valgus alignment among 237 subjects, and the development of osteoarthritis. Results from this study illustrated varus and valgus alignments to produce a fourfold increase in the risk of medial and lateral tibiofemoral osteoarthritis respectively; determined by the presence of tibiofemoral osteophytes. Similarly, this was again supported in an 18 month follow-up study, where Cahue, Dunlop, and Hayes (2004) demonstrated malalignment to influence the likelihood of patellofemoral osteoarthritis development. Varus alignment was shown to increase the risk of osteoarthritis isolated to the medial patellofemoral compartment, and a valgus alignment to the lateral patellofemoral compartment.

In the normal knee, the medial compartment carries a greater proportion of the load when compared to the lateral knee, and during gait it has been suggested that the loads across the

medial compartment are over twice the magnitude of those acting through the lateral compartment (Baliunas, Hurwitz, Ryals, Karrar, Case, Block, & Andriacchi, 2002). As such, although osteoarthritis is common in both the medial and lateral compartments of the knee joint, the prevalence of the condition is greater in the former location (Maquet, 1984).

Therefore the weight of the research suggests that the mechanical effects of lower limb malalignment on load distribution create a biological possibility that varus and valgus alignment can markedly increase the risk of debilitating conditions such as osteoarthritis development at the knee joint.

4.2.5. Knee joint stiffness.

4.2.5.1. Laxity.

Knee joint stiffness and laxity are clinical measures of knee joint behaviour, and adopt an inverse relationship (Hewett, Shultz, & Griffin, 2007). Laxity refers to the stability of a joint, and subsequently represents the degree of abnormal motion occurring at that joint (Alter, 2004). During the stance phase of gait, the force is applied externally to the knee joint to displace the tibia on the femur (Hewett et al., 2007). Defined, laxity is the amount of joint displacement at a given force, and can lead to episodes of instability of the knee joint (Lewek, Rudolph, Snyder-Mackler (2004).

According to Gali, Rigoldi, Brunner, Virji-Babul, & Giorgio (2008) the impedance of dynamic joint stabilisation associated with joint laxity has been related to increased incidence of musculoskeletal deformities. Too little stiffness at the joint may allow for excessive joint motion leading to soft tissue injury (Butler et al., 2003). Granata, Padua and Wilson (2001) demonstrated reduced knee joint stiffness and increased laxity among females during a hopping task, which could provide explanation for higher rates of knee ligamentous injuries among females. Furthermore, unsteady gait can lead to increased cautiousness during walking and running, in turn causing lower velocities and shorter stride length (Gali et al., 2008). Laxity of the knee joint is suggested to be multifactorial, resulting from such factors as

increased capsule-ligamentous laxity, altered lower extremity muscular strength, structural damage to the knee, and changes in neuromuscular control (Miura, Takasugi, Kawano, Manabe, & Iwamoto, 2009). Furthermore, excessive joint laxity can be a result of chronic injury or develop through congenital or hereditary conditions (Alter, 2004).

4.2.5.2. Joint stiffness.

Stiffness of a joint in the human body in simple terms involves the loss of normal active and passive range of motion for that joint (Copeland, Gschwend, Landi & Saffar, 1997). Range of motion is important for activities, as various degrees of movement are required for mobility (DeLisa, Gans, & Walsh, 2005). Biomechanically, knee joint stiffness can be assessed using three-dimensional gait analysis, and is defined as follows:

$$k_{joint} = \Delta M/\Delta\theta$$

Equation 4.1

where ΔM reflects change in joint moment and $\Delta\theta$ reflects change in joint angle (Farley et al., 1998). Therefore in theoretical terms, greater stiffness of the knee joint in the sagittal plane will result from reduced knee excursion solely or in conjunction with higher knee moments (Dixon, Hinman, Creaby, Kemp & Crossley, 2010).

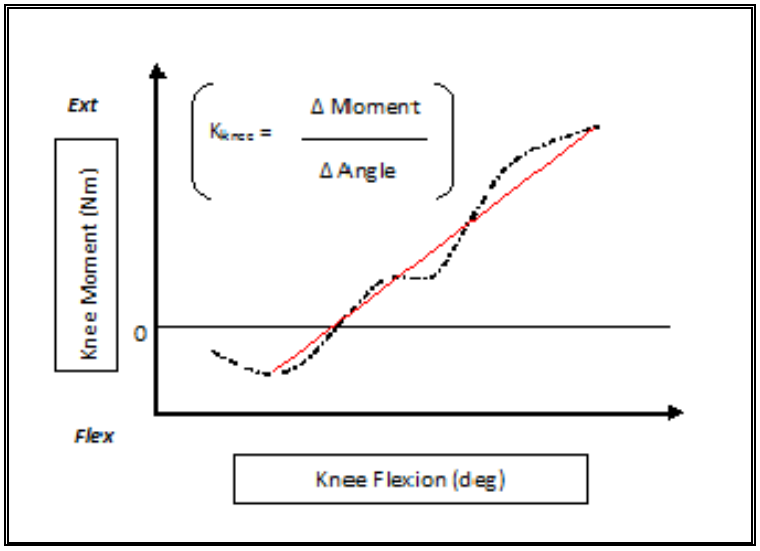


Figure 4.2. Image of knee joint stiffness in the sagittal plane of movement. Torsional stiffness calculated as the gradient of the best fit line through the moment-angle curve, from maximum flexion moment to extension moment. Graph created from sample data.

As illustrated in Figure 4.2, reduced magnitude of joint angle or increased joint moment will cause an increase in the torsional stiffness measured. However, previous research has suggested additional explanations for changes in dynamic joint stiffness. Although the regulation of the spring-mass system of the leg is predominantly controlled by the central nervous system, the situation at the joint is somewhat different (Kuitunen et al., 2002). The stiffness of a single joint is influenced by several variables. One of the most frequently cited factors affecting stiffness is the co-contraction of antagonist and agonist muscles, and according to Zeni and Higginson (2009) high antagonistic muscle activity can underlie an increase in joint stiffness. According to Hobara et al., (2009), muscle morphology may also affect stiffness, and when compared to the ankle joint, the muscles surrounding the knee have longer muscle fibers, larger volume, and greater cross sectional area. These characteristics may produce greater joint moments, increasing stiffness, which could additionally provide explanation for the greater contribution that knee joint stiffness has to overall leg stiffness (Hobara et al., 2009). When muscles are shorter than their ideal length they are at a biomechanical disadvantage when it comes to generating force. Dixon et al (2010) later

supported this notion suggesting that active elements at the joint including muscle strength, activation and co-contraction may all influence joint stiffness. Furthermore, knee stiffness has been associated with viscoelastic changes in the joint, changes in articular cartilage structure, osteophytes formation, and thickening and stiffening of supporting ligaments (Oatis et al., 2006). Finally, although possibly not a direct cause, it is possible that stature and body mass index also have an effect on changes in stiffness at the joints (Oatis et al., 2006).

The magnitude of stiffness at the joints has previously been reported to increase with the demands of the activity undertaken, and changes in gait pattern could therefore be related to the magnitude of stiffness at the joints (Butler et al., 2003). According to Butler et al., (2003) as the body is exposed to increased forces; greater resistance to movement is needed to maintain controlled movements. Changes in stride length and walking speed have been shown to alter joint angles and moments, therefore affecting stiffness (Zeni & Higginson, 2009), and it has been shown that for a given velocity, a decrease in stride length is associated with increase in stiffness (Farley & Gonzalez, 1996). This is important for performance, as although increasing stride length may improve performance, there may be a negative side affect of increased stiffness which in turn could consequently decrease running velocity. This was supported by Kuitunen, Komi and Kyrolainen (2002) who reported significant increases in knee stiffness with increase in sprinting velocity. Additionally, studies have previously suggested that changes in torsional stiffness in the sagittal plane may be related to foot strike pattern during gait (Butler et al., 2003). As highlighted previously, in their study of tibial shock in forefoot and rearfoot striking runners, Laughton, McClay Davis, and Hamill (2003) assessed ankle, knee and leg stiffness to provide additional information on shock attenuation among the two groups. Results illustrated that among the forefoot strikers, knee flexion excursion was lower than the rearfoot group, indicative of higher knee stiffness when striking with the forefoot.

4.2.5.3. Methods of assessing stiffness.

Joint stiffness can be identified as either the symptom of pain on moving a joint, or the physical sign of reduced range of joint motion. Patients will recognise stiffness using the first

method; however in the clinical setting the final method is preferred by practitioners to provide an objective measure of joint stiffness. Different methods of assessing joint stiffness are likely to produce different results, and as such caution is warranted when making comparisons between results from non uniform methods (Butler et al., 2003).

As illustrated in Figure 4.2, knee joint stiffness can also be biomechanically determined by calculating the gradient of the regression line through joint moment against joint angle data (Butler et al., 2003). This method is made possible through the use of three-dimensional gait analysis, and although often not applicable to the clinical setting, it provides a reliable objective measure of stiffness ideal for research comparisons.

One of the most common methods of clinically assessing stiffness is through the use of the self-report tool (Oatis, Wolff & Lennon, 2006). The Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) is a survey comprising 24 separate questions, 5 addressing pain, 17 addressing function, and 2 addressing stiffness. Within this survey, stiffness is identified as restriction or slowness in the ease with which the joint can be moved. Although this tool has been validated and is often used when assessing treatment outcomes, the test-retest reliability of the WOMAC scale is low (Bellamy, Buchanan, Goldsmith, Campbell & Stitt, 1988). Additionally, it is limited in its ability to specify solely an individual's complaint of stiffness (Oatis et al., 2006). According to Dixon et al (2010) stiffness is difficult to both define and quantify using the self report assessment, as substantial individual variability may exist in the interpretation of what constitutes stiffness. Therefore although easy to carry out, the reliability of a self report tool is limited, and it is suggested that future research should look to develop an alternative clinical tool to augment self-reported results (Dixon et al., 2010).

4.2.3.4. Knee joint stiffness, osteoarthritis and injuries.

Stiffness is a common complaint of patients with osteoarthritis. Clinically, knee joint stiffness begins with inflammation, immobility, and quadriceps weakness (Norris, 2004). Patients presenting knee stiffness are unable to gain full extension at the knee joint, and patella glide is restricted. Ranges of motion deficits are a well known sequelae of osteoarthritis. Knee

osteoarthritis classically involves extension lag, but flexion may also be limited. The pathophysiology of range of motion deficits is most probably multifactorial in origin including articular changes within the joint as well as shortening of myotendinous structures in areas of pain or weakness. Decreased range of motion is often found at other joints within the same limb and even in the contra lateral limb.

Reduced knee range of motion and increased knee joint moments associated with knee joint stiffness could increase the rate of loading during gait, and subsequently increase risk of knee joint injuries (Butler et al., 2003). As such, some studies have investigated knee joint stiffness as a possible risk factor. In one prospective study, although stiffness was not directly measured, a group of non injured runners were shown to run with decreased ground reaction forces, but similar knee flexion angles compared to a group of runners that sustained an injury, suggesting the latter group had higher joint moments and subsequent stiffness values (Hewett et al., 1996). A similar finding was demonstrated in a retrospective study, where Williams (2003) showed a group of high arched runners with increased leg stiffness to sustain an increased number of injuries when compared to low arched runners with reduced stiffness levels. In contrast, Granata and colleagues suggested low levels of stiffness are a risk factor for injury, due to increased instability of the joint (Granata, Padua, & Wilson, 2001).

As illustrated, although knee joint stiffness has been linked with both osteoarthritis and injuries, the results in the literature are not conclusive. When investigating the association with osteoarthritis, increased research is required into the role of stiffness as a cause or consequence of the condition. According to Butler et al (2003), an increased number of studies are required to substantiate the association between joint stiffness and injuries and osteoarthritis, with a prospective style enabling the determination of stiffness as a risk factor or consequence.

Dynamic knee joint stiffness describes the interaction between changes in joint moment and changes in joint angle during the stance phase of running and walking gait (Zeni & Higginson, 2009). Knee joint stiffness has previously been related to degenerative changes, and as such, it has often been characterised as one of the 6 criteria for clinical diagnosis of osteoarthritis (Gignac, Davis, Hawker, Wright, Mahomed, & Fortin, 2006). Based on the review of literature presented in Chapter 2 (2.7), within this investigation, knee joint stiffness during running is investigated, in its association with knee joint loading and potential osteoarthritis and injury risk. According to Farley and colleagues, joint stiffness is a

quantitative measure of the resistance that muscles and soft tissues provide during joint and limb movement (Farley, Houdijk, Van Strien, & Louie, 1998). As such, increases in muscle activation are related to increases in joint stiffness (Gerritsen, van der Bogert, & Nigg, 1995).

In terms of gait, analysis of dynamic knee joint stiffness may provide insight into the different strategies used to overcome instability of the knee joint. Previously, it has been shown that women demonstrate lower values of joint stiffness, and consequently higher levels of instability compared to males (Hsu, Fisk, Yamamoto, Debski & Woo, 2006). This suggests that women may have lower muscle strength compared to males; a potential risk factor for the development of degenerative conditions and injuries at the knee joint (Cammarata, & Dhaher, 2008). The association between joint stiffness and variables of gait was further supported by Farley and colleagues who suggested that foot strike pattern directly affects stiffness of the ankle and knee joints, although this was shown among participants during hopping movements (Farley et al., 1998). In addition, it has previously been postulated that increased stiffness is associated with decreased knee range of motion and increased peak forces, leading to increased loading rates (Butler et al., 2003). However the direct relationship between knee joint stiffness and loading rates is not conclusive during gait, and it has previously been shown that stiffness did not contribute to rate of loading of ground reaction force during walking (Dixon, Hinman, Creaby, Kemp & Crossley, 2010). Dixon and colleagues also investigated the association between knee joint stiffness and the knee external adductor moment among osteoarthritis patients during walking, and found no association between the variables (Dixon, et al., 2010). However, as is apparent in one investigation, and among many others, the association between knee joint stiffness and features of gait that may lead to osteoarthritis has commonly been examined among patients already diagnosed with the condition. Therefore the affect of structural changes associated with the condition may influence stiffness data and gait variables (Dixon, et al., 2010). Furthermore, the association between knee joint stiffness and osteoarthritis risk has previously been an indirect one, where loading rates have often been used as representatives of joint stiffness (Butler et al., 2003).

Although the direct association between knee joint stiffness and injury rates is not well established, knowledge of knee joint stiffness among mature females may provide insight into the differences displayed in running gait, and the high loading rates among this group. Furthermore, high magnitudes of knee joint stiffness may support the theory that it is the

running gait of mature females that predisposes this group to degenerative joint changes and injuries among this group.

4.2.6. Quadriceps strength.

4.2.6.1. Factors influencing muscle strength.

Muscular strength is an accepted component of health related fitness, and largely affects human ability to perform tasks of daily living without undue physical stress and fatigue. Strength is formed from an interaction between the musculoskeletal system and systems that provide metabolic, hormonal and neurologic support (Frontera, 1999).Whether muscles contract for strength or endurance, many factors influence the development of active forces in the muscles (Rybski, 2004). The main determining factor influencing muscle contraction is the physiological makeup of the muscle, including muscle fiber type, arrangement and size (Rybski, 2004). However other factors including age, gender, health status, psychological well being, and level of physical activity also present an immediate and direct impact on muscle function (Frontera, 1999). Additionally, joint position and type of muscular contraction affect the torque produced, however during assessment these are generally controlled for (Clarkson, 1999). As two of the factors involved in this thesis, both age and gender effects on muscle strength are assessed in greater depth.

4.2.6.1.1. Gender.

According to Rybski (2004), females are approximately 40 to 50% weaker in the upper body and 20 to 30% weaker in the lower body when compared to males of similar age. Many factors are suggested to contribute to this discrepancy, giving men a mechanical advantage over women, enabling them to handle more weight and generate more power. The large gender difference in upper body strength has previously been accounted for by the greater

proportion of tissue distributed in the upper body among males (Miller, MacDougall, Tarnopolsky and Sale, 1993). One additional gender difference includes the shorter extremities and subsequent length of bones in the female skeleton. Finally, the reduction in strength in the lower extremity has been previously related to the wider pelvis relative to length of the femur inherent in the female body. This produces an increased Q angle, which has been shown to negatively affect optimal muscle firing in lower extremities, suggesting the presence of a relationship between muscle strength and biomechanical alignment (Holloway & Baechle, 1990). This effect was further supported by Kumar (2004) who suggested that instantaneous muscle length, influenced by joint angles and body position, is a crucial determinant of strength. In addition, according to Miller et al., (1993) women were shown to display 30% smaller muscle cross sectional area for the vastas lateralis muscle when compared to aged matched males, affecting the absolute muscle strength achieved.

4.2.6.1.2. Age.

The ability of the muscles to produce torques of high magnitude increases from birth to a maximum point between 20 and 30 years of age (Clarkson, 1999). From the age of 65 decreases in isokinetic muscle strength occur at a rate of approximately 2.5% per year, depending on specific muscle group and contraction velocity (Chandler & Brown, 2008). This progressive loss of muscle mass with increasing age is termed sarcopenia (Chandler & Brown, 2008). This notion is supported by Robergs and Keteyian (2003) who suggested that muscle strength begins to decline around the age of 40 years, with an accelerated decline after 60 years. According to Larson, Grimby and Karlsson (1979), it is reasonable to assume that an important part of the changes in muscle strength with age takes place in the muscle tissue itself due to muscle atrophy (decrease in the mass of the muscle). Clarkson (1999) suggested a decrease in the size and number of muscle fibers, an increase in connective tissue and fat, and a decrease in the respiratory capacity all contribute to an overall decrease in strength of muscles as age progresses. Additionally, as age increases, the tendency towards a sedentary lifestyle increases, and decreased muscle mass has often been attributed to such changes in lifestyle and decreased use of the musculoskeletal system (Robergs and Keteyian, 2003).

Several studies have investigated this relationship between muscular strength and advancing age. In one early study, Larsson et al (1979) investigated both dynamic and isometric quadriceps strength characteristics of participants aged between 11 and 70 years. Results of this study indicated strength to increase from 11 to 30 years, remain constant into the fifth decade, and then decline from 50 years onwards. Muscle biopsy results further illustrated a decrease in type II muscle fiber area, and type II muscle fiber atrophy in subjects from 50 years and above. Therefore this supports the suggestion of decreased muscle strength with age. However, it is noted that this study only investigated male subjects, and all subjects were categorised with a low physical activity level (Larsson et al., 1979). A later study further investigated this relationship between age and strength in 654 women and men aged 20 to 93 years (Lindle, Metter, Lynch, Fleg, Fozard, Tobin, Roy, & Hurley, 1997). Differences in isometric, concentric and eccentric torque of the quadriceps muscles were assessed, with results once again showing significant age-related declines (8-10% per decade) in knee extensor strength across both sexes. Significant decreases were also shown in the muscle quality (strength per unit of muscle) or concentric knee extensor torque in both sexes with increasing age. However it was noted here that only men presented a decline in the muscle quality of the peak eccentric knee extensor torque. Additionally, age-related declines in isometric and concentric strength began in the forth decade for both sexes, however the age-related decline for eccentric muscle strength began a decade later for women. This suggests older females have a greater capacity to store and utilise elastic energy; a function of eccentric muscle contraction (Lindle et al., 1997). However although these results once again support the theory of age-related decline in muscle strength, it was documented that less than 1% of participants took part in any regular resistive activity, and no record was made of participation in regular physical exercise.

Murray, Duthie, Gambert, Sepic and Mollinger (1985) lent support to this trend of age-related strength decline in an investigation of knee muscle strength among women. Results from this investigation illustrated the isometric and isokinetic recorded muscle strength of women aged up to 86 years to reach only 56% that of women aged 20 years. Additionally, Young, Stokes and Crowe (1984) showed quadriceps strength of women in their 70s to be 35% weaker than that of women in their 20s, with a 33% reduction in quadriceps cross sectional area. This investigation showed a correlation between strength and area, and suggested that there is no difference in the intrinsic strength of the quadriceps muscles in women aged between 30 and 80 years (Young et al., 1984). However a later study by Pearson, Bassey and Bendall (1985)

suggested alternative results and showed age to have a significant negative correlation with strength of the triceps surae muscle, and strength per cross sectional area of the calf muscles. Therefore, changes within the muscle must account for decreases in strength (Pearson et al., 1985). However it is noted that the population investigated were females aged between 65 and 90 years, and as the approach was not longitudinal, significant strength changes may have already occurred at an earlier age.

Therefore in light of the research it seems apparent that a clear decline in strength occurs with age among females. However, the effect of regular physical activity on this strength decline among the more mature groups has received little investigation. It seems important to investigate quadriceps strength among females of advancing age compared with younger females that take part in similar levels of regular physical activity.

4.2.6.2. Quadriceps strength, gait and injury.

Due to the close association between biomechanical movement patterns and the anatomy and physiology of the human musculoskeletal system, it is likely that strength of the muscles in the lower limb could affect running gait. Although not equivocal, some evidence supports an association between reduced muscle strength and injury risk (Bartlett, 1999). Experimental data combined with conventional knowledge lead to the theory that the muscles in the human body serve as the main shock absorbers during gait (Mikesky, Meyer, & Thompson, 2000). According to Gage (1993), muscles absorb force through eccentric (lengthening) contraction. As such, muscle weakness or imbalance can often lead to compensatory changes in movement patterns, which can in turn increase the stress on segments elsewhere in the kinetic chain during gait (Bartlett, 2002). This theory is supported by Lewek and colleagues, who suggested an association between reduced strength of the quadriceps muscles, and reduced range of motion at the knee joint during walking and running (Lewek, Rudolph, Axe & Snyder-Mackler, 2002). The quadriceps muscles, involving the vastas medialis and vastas lateralis are recognised as major contributors to patellofemoral joint function; they act to facilitate restraint of the patella in medial and lateral directions respectively. Abnormalities or weaknesses of these muscles have been linked to patellofemoral pathologies including

osteoarthritis of the knee joint (Felson, Naimark, Anderson, Kazis, Castelli & Meenan, 1987). Insufficient strength of the quadriceps muscles could interfere with the shock attenuating function of the knee joint during stance, leading to degenerative changes to occur (Lewek et al., 2002). Similarly, among females, reduced quadriceps strength has been associated with increased loading rate during the stance phase of walking gait (Mikesky et al., 2000). This investigation found that the strength-trained group were on average 29% stronger, and had significantly lower rates of loading, decreasing the risk to the long-term joint integrity (Mikesky et al., 2000). Mikesky and colleagues related this to an earlier study by Radin and co-workers, who found an association between rates of loading and knee pain, and theorised that individuals with high rates of loading had lower muscle strength (Radin, Yang, Reigger, Kish, & O'Connor, 1991). Therefore, the proposed association between loading rates and muscle strength highlights the importance of strength assessments among mature female runners. It is possible that lower quadriceps strength among this group could be a factor in the high loading rates, and relative risk of injuries and osteoarthritis at the knee joint.

4.2.6.3. Assessing Quadriceps Femoris strength.

Muscular strength is specific to the muscle group being tested, and as such no single test exists to assess total body muscular strength or endurance (Robergs & Keteyian, 2003). As described above, muscular strength refers to the ability to exert maximum muscular force statically or dynamically. The measurement of the quadriceps muscle torque at the knee joint using isokinetic dynamometry is very useful in providing an insight into muscle function and in obtaining muscle performance data for various modelling purposes. (Barlett, 2002). Isokinetic testing involves an electromechanical resistance instrument containing a speed controlling mechanism that enables constant velocity with the application of a force (McArdle, Katch & Katch, 2003). This enables an individual to generate maximum force or a percentage of maximum effort, through the full range of motion at a pre-established velocity of limb movement (McArdle et al., 2003). A microprocessor within the dynamometer acts to continually monitor the immediate level of applied force and the voltage output is transferred directly to a computer for almost instantaneous feedback. When considering methods of assessing muscle strength, it is however important to bear in mind the possible influence of

the nuances of the testing procedures used, including degree of practice, level of motivation, and level of understanding of the purpose and nature of the test (De Ste Croix, 2007).

4.2.7. The influence of footwear.

As mature females have been shown to exhibit patterns of gait that differentiate them from a younger group, it was deemed important to investigate methods to alter this pattern of gait in attempt to reduce the injury rates. Within the first study of this thesis, all subjects wore their own footwear. According to Froncioni (2006), running trainers are a high-tech device purpose-designed to protect the runner from injury and improve performance; however two out of every three runners are still side-lined every year with a running related injury. The interaction between the sport shoe and the runner has therefore previously been subject to much speculation, and has often been considered to influence impact forces, loading, and foot pronation (Nigg, 2001). As such, mechanisms of cushioning and motion control properties of running shoes have been subject to extensive research, in an attempt to determine adequate strategies in injury prevention (Nigg, 2001).

Although limited, the capacity to reduce such injuries often involves training advice and prescription of suitable footwear (Richards, et al., 2009). Footwear is essential equipment for runners (Cheung & Ng, 2007), and in recent years, an abundance of biomechanical research has focussed on the properties of footwear. This has taken place with a variety of functions, including optimising sporting performance, reducing injuries, and recently enhancing muscle tone and weight loss (Fit Flops Inc). Certain shoes are specifically designed with technologies aiming to increase cushioning, and reduce impact forces during running. Most frequently, shock absorbing properties within running shoes involve low stiffness foams and gels, and viscoelastic materials which reduce the acceleration of the heel at heelstrike (Noe, Voto, Hoffman, Askew, & Gradisar, 1993). These footwear types generally fall into the bracket of “neutral” or “cushioning” shoes, and do not contain additional support to control the alignment of the foot (Drenth, 2011). As such, the inclusion of a neutral shoe in the investigation of mature female runners will negate the effect of different footwear, in a comparison between lower limb kinematics of a mature and young group.

4.2.7.1. Cushioning properties in footwear.

By virtue of their anatomical position, the foot and ankle create a dynamic link with the ground. Recent advances in biomechanical technology have enabled a quantitative evaluation of the foot and ankle function during the process of running (Shariatmadari, English & Rothwell, 2010). In general, the majority of footwear research has focussed on rearfoot strikers, and aims to modify forces and control motion of the calcaneus immediately following heel strike (Spurgeon, 2005). Over the course of one mile, an individual will make approximately 1600 foot strikes (Spurgeon, 2005). Throughout running, impact forces are transmitted to the lower extremity with every stance phase of gait. During this phase, at heel strike the foot accommodates both a vertical impact force and a horizontal shearing force that creates friction (Norris, 2004). Anatomically, the viscoelastic properties of the tissues in the heel provide significant shock absorption, however impact forces and rate of loading have still been linked with the development of overuse conditions (primarily osteoarthritis) and injuries such as stress fractures, plantar fasciitis and knee pain (James, Bates & Osternig, 1978; Ferber, McClay-Davis, Hamill, Pollard, & McKeown, 2002). A wide variety of materials and design concepts have been added to footwear in attempt to produce shoes that will absorb a large proportion of the impact energy attenuated during running gait (McNair & Marshall, 1994). This is generally referred to as the cushioning property of running footwear, and has been shown to reduce impact force at heel strike (Subotnick, 1983). Overall, the effect of footwear on the vertical ground reaction force peak and loading rates have elicited varied results.

4.2.7.1.1. Footwear and Loading rates.

As a variable of gait that has been linked with injuries within the muscle strength and knee joint stiffness literature, the effect of footwear on the loading rate during running is discussed. Although the effect of loading rate on injury risk and development of osteoarthritis has been subject to much speculation (Gill and O'Connor, 2003), many footwear companies have

focussed on shock-attenuating properties within shoes, resulting in a huge industry in running shoe design (Kirtley, 2006).

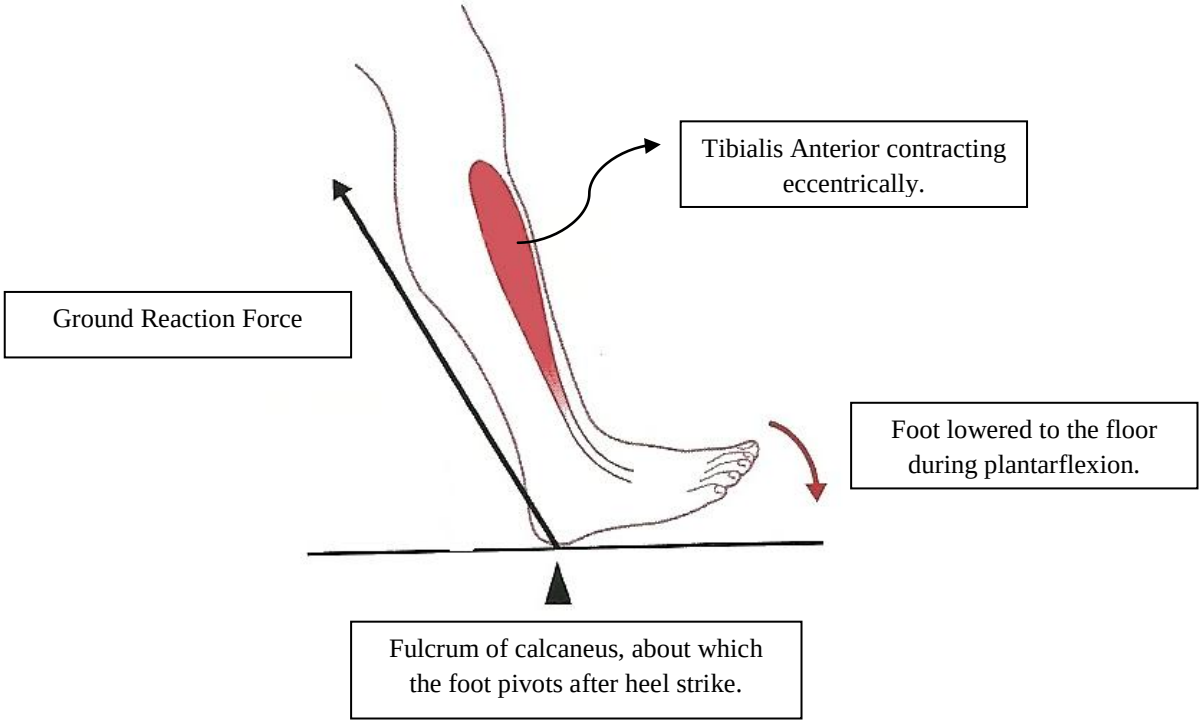


Figure 4.3. Illustration of the foot-ankle complex at heel strike, indicating the ground reaction force acting posterior to the ankle joint, and the action of the tibialis anterior muscle. Image adapted from Kirtley (2006).

During gait, the initial movement following heel strike involves plantarflexion of the ankle joint, which pivots around the calcaneus, and enables absorption of the impact forces (Figure 4.3) (Kirtley, 2006). This is achieved through three mechanisms; the fat pad of the heel, which absorbs the shock by providing a layer of anatomical cushioning, the eccentric contraction of the dorsiflexors (tibialis anterior muscle), and pronation of the subtalar joint

(Kirtley, 2006). This latter mechanism unlocks the transverse tarsal joint, and enables the flexible foot to adjust to alterations in the ground surface, and reduces the shock travelling up to the proximal joints of the lower limb (Norris, 2004). As such, it is proposed that a motion control shoe, designed to reduce pronation of the subtalar joint, may limit this mechanism, and result in increased loading rates during running in the early stage of stance. This proposition was partially supported by Lee and colleagues, who investigated the effect of motion control and neutral shoes on both treadmill and over ground running. Within this study it was shown that during over ground running, the motion control shoe increased the maximum loading rate by an average of 2.8 % compared to the neutral shoe, although the significance of this difference was not noted (Lee, Lafortune, & Valiant, 2007). Butler and colleagues showed a higher rate of loading among high arched runners compared with low arched runners, linking increased loading rates with reduced subtalar joint pronation (Butler, Hamill, & Davis, 2007). Furthermore, the inclusion of a motion control shoe was shown to significantly increase instantaneous loading rate by 11.8 body weights per second when compared to a neutral cushioned shoe. This was also supported in an earlier study by Perry and Lafortune, who illustrated a motion control wedge insert to reduce rearfoot eversion, but increase vertical loading rate when compared to the neutral shoe (Perry & Lafortune, 1995). Within the current study, it will therefore be important to assess any changes in loading rate with the inclusion of a motion control shoes among the mature female runners.

According to Lees & McCullagh (1984), viscoelastic shock absorbing shoe insoles have been shown to produce reductions in vertical peak impact force and loading rates among runners. Similarly, Clarke and colleagues illustrated an increase in cushioning of running shoes to significantly increase the time to peak impact force, although no change was shown in the magnitude. Therefore the cushioning aspect of the shoes appears to alter the loading rate of impact force (Clarke, Frederick & Cooper, 1983). This has also been shown during walking trials where softer insoles can cushion the impact forces at heel strike (Lafortune & Hennig, 1992). In contrast, Nigg and colleagues demonstrated no significant differences in ground reaction force variables when viscoelasctic shock absorbing insoles were added to running shoes (Nigg, Herzog & Read, 1988). It is however noted that this latter study only investigated group reactions to insoles, and individual variations may have been overlooked.

4.2.7.2. The motion control effect.

According to McPoil (2000), footwear during gait has four main functions, each developed to enhance the prevention of running injuries. Shoes act to protect the plantar surface of the foot, provide traction between the foot and the ground, control motion, and aid attenuation of impact forces during activity. However of the four main functions, it is the mainstream technology of motion control design that has been the subject of the majority of recent research interest (McPoil, 2000).

Evidence exists to suggest that footwear may be able to alter motion of the foot during walking and running (Rose et al., 2011). The term rearfoot control is based on the ability of a shoe to limit the movement of the rearfoot following heel strike (Clarke, Frederick & Hamill, 1983). Although a certain amount of rearfoot motion is essential during gait, the current concept of the motion control shoe is to reduce the chance of excessive motion, which is commonly associated with injuries (section 3.1.3.2). It has been suggested that excessive rearfoot eversion during running and associated increased internal rotation of the lower limb causes a change in load distribution, and a relative increase in load acting at certain areas of the lower extremity (Taunton et al., 2002).

Several investigations into rearfoot and forefoot motion during running have therefore occurred to determine the desired amount of control that should be imposed by the shoe on the foot. In general, the motion control design targets to reduce excessive rearfoot eversion and subsequent internal rotation of the lower leg and knee, during running, through altering deformation rates and geometry between the medial and lateral sides of the shoe, (Cheung & Ng, 2007). Increased density and height is therefore added to the medial sole of motion control shoes, to support the medial longitudinal arch and reduce the movements associated with subtalar joint pronation.

In a study of footwear design during running, Clarke and colleagues illustrated that shoes containing both medium and hard midsoles (Shore 35 and 45) significantly reduced maximum rearfoot eversion and total rearfoot movement compared to those with softer midsoles (Shore 25) (Clarke et al., 1983). Hamill, Freedson, Boda and Reichsman (1988) supported this in a later investigation into differences in rearfoot movement among women wearing motion control shoes and racing flats. Results showed an average 42% increase in maximum rearfoot eversion in the flats when compared to shoes with greater medial support.

This theory was later supported by Cheung and co-workers who showed increased medial support in motion control shoes to significantly reduce average rearfoot eversion by 4° among a group of 28 female runners (23.5±6.8 years) (Cheung & Ng, 2007). In terms of injury rates and reduction methods, Knapit and colleagues examined the effect of matching the support in shoes to foot type and pronation level, and showed 50% reduction in injuries to the lower extremity and the back over a six month period (Knapik, Feltwell, Canham-Chervek, & 2001). However it is noted that the mechanisms behind the injury reductions were not detailed and therefore cannot be commented on.

According to Morio and colleagues, when compared to barefoot locomotion, footwear causes reductions in the rearfoot eversion angle and velocity; during running, the total range of rearfoot eversion was reduced by 20 % in the footwear condition (Morio, Lake, Gueguen, Rao & Baly, 2009). The reduction of a heel counter (as described earlier) was investigated by Jorgensen (1990) among a group of asymptomatic heel-strike runners. Results showed that when the firm heel counter was removed, the electromagnetic activity of the triceps surae muscles occurred significantly earlier than when the heel counter was in place, suggesting an increased action of the triceps surae muscle group to maintain stability in the former condition (Jorgensen, 1990). Butler and colleagues also showed the effect of motion control shoes on the musculoskeletal system, and illustrated that when compared to a cushioned (neutral) shoe, a stability (motion control) shoe significantly decreased the tibial internal rotation angle among a group of low arched runners (Butler, Hamill, & Davis, 2007).This effect suggests that motion control shoes may be a method to reduce the patellofemoral joint strain that has been associated with low arched runners (Williams, McClay, & Hamill, 2001).

These researchers support the overall theory that increasing support in the medial sole of the shoe controls for subtalar joint pronation. However, speculation still exists as to the correct location of support along the medial sole to best control for excessive frontal plane movement of the rearfoot. A review of early literature has revealed a range of published studies with varying results. One early study by Shaw suggested that forefoot support is adequate to control for pronation during gait; a theory later supported by Ramig and colleagues (Shaw, 1975; Ramig, Shadle, & Watkins, 1980). Novick and Kelley later disputed this, suggesting that both rearfoot and forefoot support in the shoe bed was required to control for pronation during gait, and Subotnick showed that rearfoot posting alone is sufficient to control for pronation, including forefoot motion, during gait (Novick & Kelley, 1990; Subotnick, 1983).

Although the former results were shown during walking trials, this latter result supports the theory that rearfoot eversion is the dominant component of pronation. Johanson and colleagues performed a later study which supported the previous two theories, and suggested that due to the combination of movements that produce subtalar pronation, ideal control is achieved through heel focussed medial support, or combined heel through to forefoot support (Johanson et al., 1994). These authors suggest that posting at the medial arch alone, as is found in many motion control shoes, does not adequately control for excessive subtalar joint pronation (Johanson et al., 1994).

Although many studies support the use of motion control shoes to reduce rearfoot motion, it is important to note that a certain degree of rearfoot eversion is due to the natural adaptive movement of the subtalar and mid-tarsal joints for shock absorption and thus is beneficial to gait (Nordin & Frankel, 2001). Furthermore, it is important to note that the predominant feature of pronation is rearfoot eversion, and as described above, in the general motion control shoe, the support tends to be centred on the medial arch, supporting the foot later in stance (Johanson et al., 1994). Although there have been studies that have supported the use of motion control shoes to alter tibiocalcaneal kinematics during running, other researchers have shown no effect (Stacoff et al., 2000; Butler et al., 2007). As suggested by Clarke, the efficacy of motion control shoes is highly dependent on the foot form and structure, and many studies have not controlled for heterogeneity (Clarke et al., 1983). Therefore it is suggested that equivocal results may be due in part to variations in support design, the materials used, and the level of posting within the shoe. Although evidence regarding the use of motion control shoes to reduce rearfoot eversion is not without dispute, the promising results produced in some studies have caused footwear adaptations to often be implemented in injury prevention and rehabilitation. As such, the effect of a motion control shoe on lower limb kinematics will be assessed among mature female runners.

4.2.7.3. Footwear and the knee external adductor moment.

As discussed, the motion control aspect of footwear is implemented to control excessive subtalar joint pronation and subsequent internal rotation of the lower leg and knee. In

contrast, other researchers have investigated the effect of footwear modifications to reduce the knee external adductor moment that has been associated with knee joint osteoarthritis (section 3.4.5) (Hunt et al., 2006). This biomechanical variable occurs as the ground reaction force and centre of pressure act medially to the knee joint centre causing an adductor moment to act at the knee (Hunt et al., 2006). This is compounded with an increased varus alignment as the moment arm from the knee joint centre to the line of ground reaction force increases, causing an increased load experienced by the medial relative to the lateral compartment of the knee joint (Figure 4.4). Over time this increased loading leads to a general degeneration of the medial tibiofemoral articular cartilage, and the development of certain conditions such as osteoarthritis (Hunt et al., 2006). According to Hunt and colleagues, the magnitude of this moment is highly dependent on the length of the moment arm described (Hunt et al., 2006). As such, methods to reduce this moment arm are often centred around altering the position of the foot on the ground, to reduce the varus position of the knee, and move the centre of pressure laterally on the foot (Kakihana et al., 2005).

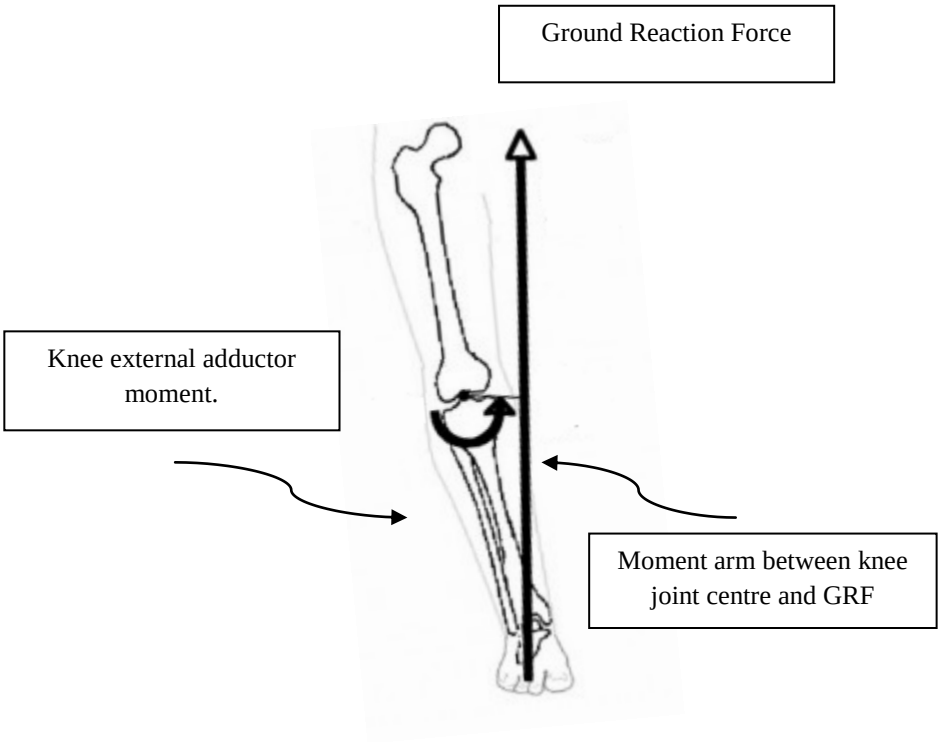


Figure 4.4. Diagram of the knee external adductor moment acting as a combination of the ground reaction force and the distance to the knee joint centre. Adapted from Russell et al., 2010.

Current interventions to reduce this knee external adductor moment and subsequent loading of the medial knee are therefore seemingly opposite to the interventions designed for rearfoot eversion. More specifically, lateral wedging has been investigated with the goal of reducing symptoms and progression associated with medial compartment knee osteoarthritis. If proven effective in knee external adductor moment, this specific intervention could be deemed useful for future management and prevention of osteoarthritis of the knee joint. According to Crenshaw (2000), a 5° wedge on the lateral side shifts the centre of pressure laterally, reducing the external adductor moment and knee abduction (varus) positions, and offloading the medial knee. This is illustrated in the Figure below.

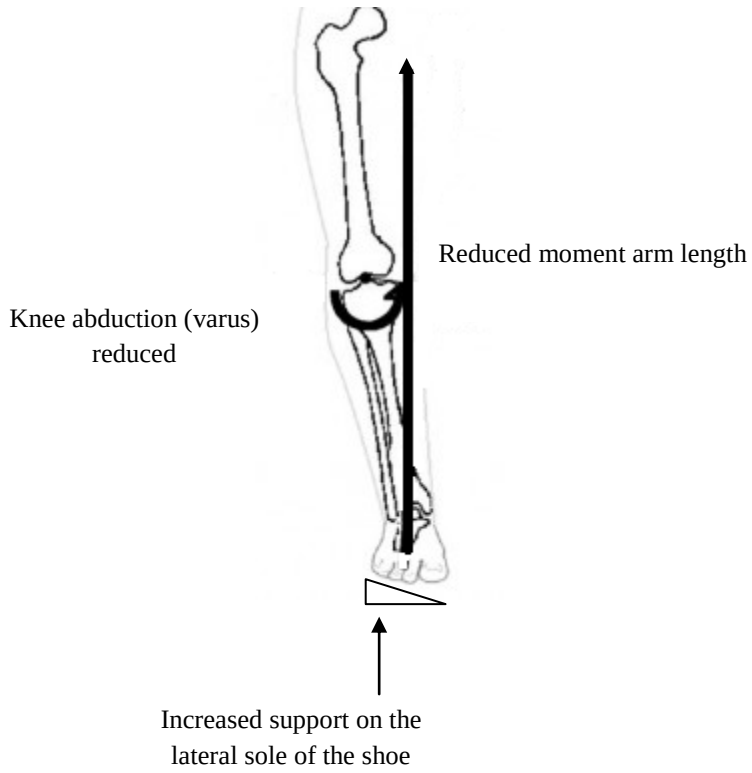


Figure 4.5. Lateral wedge added to decrease moment arm length at the knee joint, and reduce the knee external adductor moment. Adapted from Russell et al., 2010.

As such, the inclusion of increased medial sole support, common in the current motion control shoe is expected to have an opposing effect, and increase the loading at the medial knee. This theory is supported in one study that investigated the effect of a current stability (motion control) shoe and a mobility shoe (indicative of barefoot) on the knee external adductor moment (Bourgit, Millet, & Fuchslocher, 2008). Results indicated that when compared to the barefoot condition and the barefoot simulator shoe, the motion control shoe instigated a slight increase (8%) in the external adductor moment. However these results were demonstrated among a group of symptomatic OA patients, and the gait adaptations to footwear may have been affected by pain. The effect of a motion control shoe on the knee external adductor moment of healthy runners has not been extensively researched. Based on the theories described, it is suggested that, among a group of asymptomatic females, a current motion control shoe is likely to increase the external knee adductor moment during running.

4.2.7.4. The comfort of footwear.

As well as biomechanical changes to the shoe, the concept of comfort cannot be ignored; and according to Miller and co-workers, comfort as an aspect of shoe design may be related to fatigue, injury development and performance (Miller, Nigg, & Liu, 2000). However it has been suggested that comfort is an individual response, based on a multitude of mechanical, neurophysiological, and psychological attributes unique to the individual (Miller et al., 2000). It is suggested that as malalignment and high impact forces are factors that may influence comfort, a shoe incorporating correct alignment and cushioning properties, should provide a generic level of comfort despite individual variation (Miller et al., 2000).

Among mature female runners, the influence of footwear should be considered based on the association with injuries. It is noted that footwear companies have introduced a distinction between male and female runners. However, due to the aforementioned high numbers of runners within the mature sector of the population, and a possible alteration of gait with advancing age, an investigation into the effect of footwear on the running gait of this group is deemed appropriate.

4.2.8. Aims and hypotheses

The main purpose of this study was to assess the ability of a motion control shoe to reduce specific observed differences in running biomechanics between mature and younger female runners. Three research questions were addressed:

- Is the running gait in neutral shoes significantly different for mature compared with younger females?
- Does a motion control shoe significantly alter the running biomechanics of mature and young females?
- Do physiological factors have an influence on running gait among mature females, specifically strength of the quadriceps femoris muscle?

Four hypotheses were therefore assessed:

1. Running trials performed in a consistent test shoe, marketed as a running neutral shoe, would result in greater magnitude of loading rates, peak rear foot eversion, knee internal rotation angle, knee external adductor moment, and knee joint stiffness for mature compared to younger females.
2. Quadriceps strength will be significantly lower for the mature females when compared to the younger group.
3. The motion control shoe would significantly reduce magnitudes of rearfoot eversion and knee internal rotation among females during the stance phase of running gait.
4. The motion control shoe would increase knee external adductor moment during the stance phase of gait, through an increase in the varus (abducted) position of the knees and a subsequent increase in moment arm.

4.3. Methods.

4.3.1. Participants and KOOS.

A statistical power analysis (Power and Precision, Version 2, 2000) based on the results from the previous study indicated a sample size of 12 per group would produce a power of 80%, suggesting 80% likelihood the study would yield a significant difference where present, between mature and young females. A sample report of the statistical power analysis is presented in the appendix (Appendix C). Twenty two injury free female participants were recruited on a volunteer basis for involvement in this study, and assigned to one of two groups; young and mature. The young females ranged in age from 18 to 25 years (mean 21.2 years, sd 2.1) and were recruited from the University of Exeter Sport and Health Sciences student cohort. The mature females ranged in age from 40-60 years (mean 49.7 years, sd 3.7), and were volunteers from a local women's running club (Women's Running Network). All females had a minimum of 3 years running experience and participated in at least three one hour sessions per week. Best 10-km run time was used as an indicator of standard (Table 4.1). Prior to the start of testing, each participant completed a Physical Activity Readiness Questionnaire; PAR-Q form to assess participant statistics (age, height, weight) and to ensure all females were free from any major visual or balance pathology, or debilitating condition that could hinder performance (PAR-Q, Canadian Society for Exercise Physiology, 2002). This investigation was provided ethical approval from the University of Exeter ethics committee.

Alongside this, each participant completed a KOOS questionnaire (knee osteoarthritis observation survey) to assess the presence or potential of osteoarthritis and injuries present at the knee joint (Appendix B) (Roos & Toksvig-Larsen, 2003). Results were correlated with kinematic data to investigate potential links between knee osteoarthritis and lower limb biomechanics during running gait. A summary of demographic data and KOOS scores for both groups is illustrated in Table 4.1.

Table 4.1. Demographic information for the two groups, mature and young females, including KOOS scores.

		YOUNG	MATURE
AGE (YEARS)		**21.2 (2.1)**	**49.7 (3.7)**
MASS (KG)		**60.5 (7.8)**	**58.2 (5.1)**
10-KM TIME (MINS)		**56.5**	**58.5**
KOOS SCORE		**96.5**	**90.8**
	PAIN	97.9 (1.9)	94.2 (1.9)
	SYMPTOMS	98.1 (0.9)	91.2 (1.2)
	ADL	98.7 (0.5)	92.9 (0.8)
	SPORT & REC	93.8 (0.6)	87.8 (1.5)
	QOL	94.1 (0.6)	87.9 (0.8)

4.3.2. Test running shoes.

Each participant performed 10 running trials in two running shoe models; the Adidas Supernova Sequence (motion control) and the Adidas Supernova Glide (neutral). The motion control differed from the neutral shoe in the medial section of the sole, where an increase in the density and height of the arch support was present aiming to reduce excessive pronation. Descriptions of the components inherent to each shoe are presented in table form (Table 4.2).

Table 4.2. Description of key running technologies of both Adidas shoes.

SUPERNOVA SEQUENCE	SUPERNOVA GLIDE
FORMOTION: Adapts to the ground for smoothness and comfort.	**FORMOTION**: Adapts to the ground for smoothness and comfort.
adiPRENE: Shock absorbent material that cushions and protects heel at impact.	**adiPRENE**: Shock absorbent material that cushions and protects heel at impact.
adiPRENE+: Resilient cushioning to protect forefoot and provide responsive dynamic toe-off.	**adiPRENE+**: Resilient cushioning to protect forefoot and provide responsive dynamic toe-off.
GEOFIT: Internal footwear technology enhancing fit and comfort.	**GEOFIT**: Internal footwear technology enhancing fit and comfort.
TORSION: Adaptive midfoot support.	**TORSION SYSTEM**: Lightweight arch support enabling independent movement of the forefoot and rearfoot.
adiLITE: Lightweight, Increases comfort.	**adiLITE respoEVA**: Sockliner for instep comfort and antimicrobial protection.
adiWEAR: Outsole for high-wear durability.	Women's specific outsole with larger forefoot platform and anatomically adjusted flex grooves.
PROMODERATOR: Medial device prevents over pronation.	Blown rubber outsole for lightweight grip and cushion.

4.3.3. Motion analysis and force plate assessment.

Running analysis took place in the biomechanics research laboratory. Each subject was affixed with 11 reflective markers along the right lower limb using a modified version of the model presented by Soutas-Little; greater trochanter, medial and lateral knee at the tibial plateaus, the musculo tendinous junction where the medial and lateral belly of the gastrocnemius meet the Achilles tendon, the mid tibia below the belly of the tibialis anterior, the superior and inferior calcaneus, the lateral malleolus, the third proximal head of the third metatarsal, and the distal head of the fifth metatarsal joint (Soutas-Little et al., 1987). The position of these markers illustrated the anatomical position and spatial movement of the thigh, shank, and foot during one running ground contact (Soutas-Little et al., 1987).

Three dimensional analysis of running gait occurred using an eight camera motion capture system (Vicon Peak, 120 Hz, automatic, opto-electronic system; Peak Performance Technologies, Inc., Englewood, CO.), synchronised with a single floor mounted force platform (960 Hz, AMTI, Advanced Mechanical Technology, Inc., Massachussetts). Data were synchronized within the Vicon software using initial foot strike as an automatic event detection (vertical force > 10N). The eight cameras were positioned in an oval shape focussing on the force plate. A static L-shaped frame denoting the horizontal axis of motion, and calibration wand of length 0.998 m with two reflective markers of known distance apart were used to calibrate the system. An image of the biomechanics laboratory including force plate, cameras and timing gates is displayed below (Figure 4.6).

Figure 4.6. Image of testing laboratory including force platform, timing gates, and the cameras in view.

4.3.4. Static assessment.

Prior to data collection, participants performed one single standing trial with anatomical markers attached, to provide standing joint angles for adjustment of anatomical angles during each trial. Posture was standardized with arms crossed in front of the chest, and feet shoulder width apart on the force plate. Data were collected to recognise marker orientation, relevant geometry and segment reference frames to provide the local coordinate systems. Width measurements of the forefoot (widest as observed in shoe) and ankle (from lateral to medial malleolus bones) were taken using a set of callipers. These data were then used to define the ankle joint centre and the centre line of the foot, allowing calculation of foot centre of mass for use in subsequent joint moment calculations (See previous Chapter, 3.2.3.1).

4.3.5. Analysis of running gait.

Running gait was assessed for each participant individually. Running velocity was recorded using two one meter high infrared light timing gates located along a 10 m concrete runway, situated 2.5 meters either side of the centre of the force plate, categorising each trial acceptable at 3.5 $m.s^{-1}$ ± 5% (speed determined in a pilot study). Following each trial, the force plate was set to zero. Participants were provided time to familiarise with the procedure and ensure running gait was reflective of each female's natural pattern. Participants marked the floor once a pace was set to ensure the right foot would successfully hit the centre of the force platform for each trial. Trials were accepted when running speed was within the allocated range, and the participant made full right foot contact with the force plate with no adjustment in stride. Unacceptable trials were repeated.

Mean results were collected for each subject performing 10 trials in each condition. Peak rate of loading of impact force was calculated as the rate at which impact force occurred within the first 50 ms of force generation, and presented relative to body weight. Kinematic variables of peak knee internal rotation and rearfoot eversion angles were calculated using a joint co-ordinate system approach. Knee internal rotation was calculated as the angle created as the knee rotated medially, using the knee abduction adduction vector and the position of the shank. Ankle eversion was calculated as the anatomical 180° angle between the line of the Achilles and the line of the calcaneus in the XZ reference frame. Knee joint flexion was calculated as the absolute angle infront of the knee joint (Figure 4.8). Within this investigation, knee flexion angles were positive. Finally, the transformed co-ordinates and ground reaction force data were exported from Peak into MATLAB (Version 7.0, Mathworks, Inc), and the external knee adductor moment was calculated at 120 Hz through inverse dynamics utilising a customised Matlab code, with inertia characteristics provided by Dempster (Dempster, 1955). Both ground reaction force data and moment values were then presented in relation to body weight (BW and Nm/kg). Significant differences between groups and conditions were detected through the use of a 2 way (between-within) ANOVA ($p<0.05$).

4.3.5.1. Knee joint moment arm assessment.

As described previously, the knee external adductor moment is predominantly a product of the line of the frontal ground reaction force, and moment arm between this force and the centre of the knee joint. As such, the length of moment arm was determined in order to demonstrate the influence of this variable and associated lower limb alignment, on the magnitude of the moment. As described in Chapter 2, kinematic data from the standing trial were used to obtain three-dimensional coordinates for the knee joint centre; calculated as the midpoint between the lateral and medial knee joint markers (Figure 4.7, Equation 4.2).

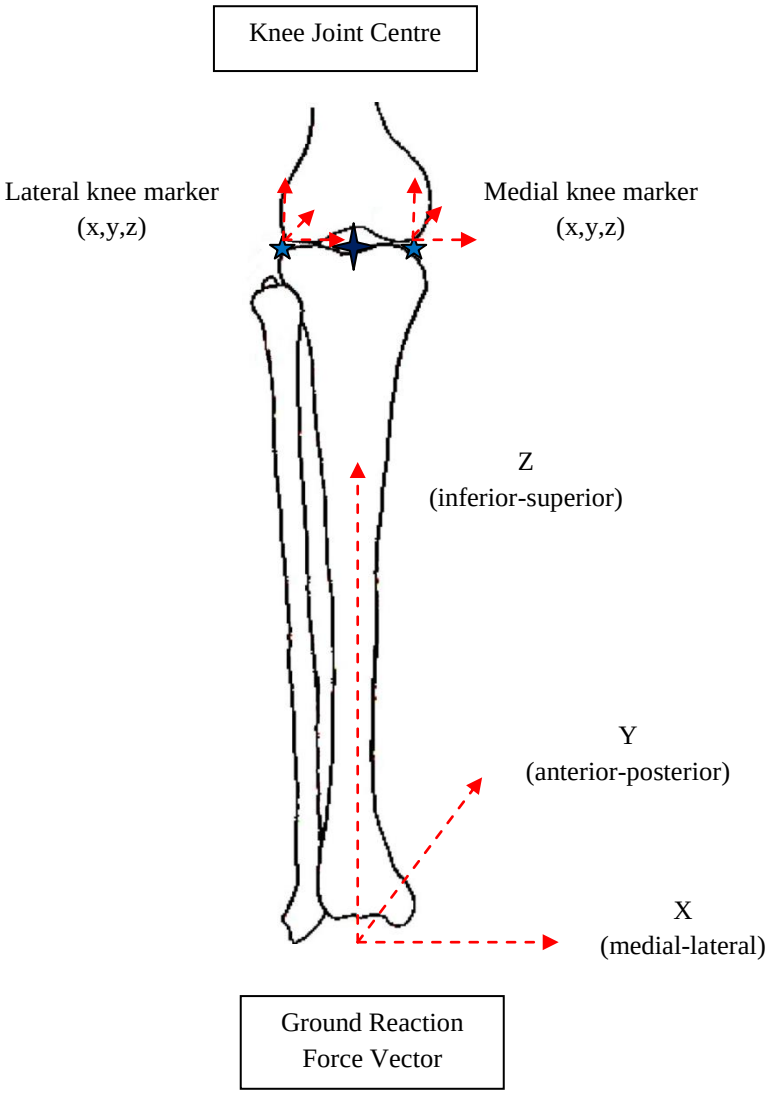


Figure 4.7. Diagram of the components for moment arm calculation at the right knee. Image of the lower leg adapted from Hunt et al., (2006).

For the purpose of the moment arm calculations, the x coordinate of the midpoint of the knee joint was calculated as follows:

$$\text{Mid knee (x)} = \text{medial knee (x)} + [0.5*(\text{lateral knee (x)} - \text{medial knee (x)})]$$

Equation 4.2

The frontal plane moment arm was calculated as the perpendicular distance between the frontal plane vector of the ground reaction force and the knee joint centre. The frontal plane ground reaction force vector was calculated as the resultant force vector of the mediolateral and vertical components of the ground reaction force as follows:

$$GRF_{frontal} = GRF_z + GRF_x$$

Equation 4.3

The distance (d) between the mid knee in the mediolateral direction and the frontal plane ground reaction force was calculated to find the perpendicular distance between the frontal plane ground reaction force and the knee joint centre.

$$d = Midknee_x * GRF_{frontal}.$$

Equation 4.4.

As such, the moment arm of the knee joint was determined, using the frontal plane coordinate of the ground reaction force ($GRF_{frontal}$), the scalar distance to the centre of pressure (d) and the knee joint centre coordinate ($Midknee_x$).

$$\text{Moment arm} = Midknee_x - (d*GRF_{frontal})$$

Equation 4.5.

Moment arm length at peak moment was calculated throughout the stance phase of gait, providing an indication of the position of the knee joint in relation to the line of ground reaction force. Negative and positive moment arms were therefore representative of the knee in an abducted and adducted position respectively.

4.3.5.2. Knee joint stiffness.

Figure 4.8 illustrates the components of knee joint stiffness; calculated as the change in sagittal plane joint angle in response to the applied joint moment during the stance phase of gait (Equation 4.7) (Dixon et al., 2010). Sample graphs are presented to show the knee angle and moment throughout the stance phase of gait (Figures 4.9 & 4.10).

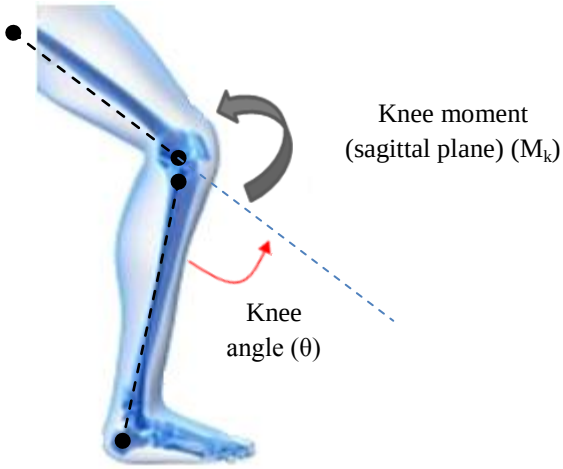


Figure 4.8. Illustration of the lower limb positioning during running, and the components of knee joint stiffness.

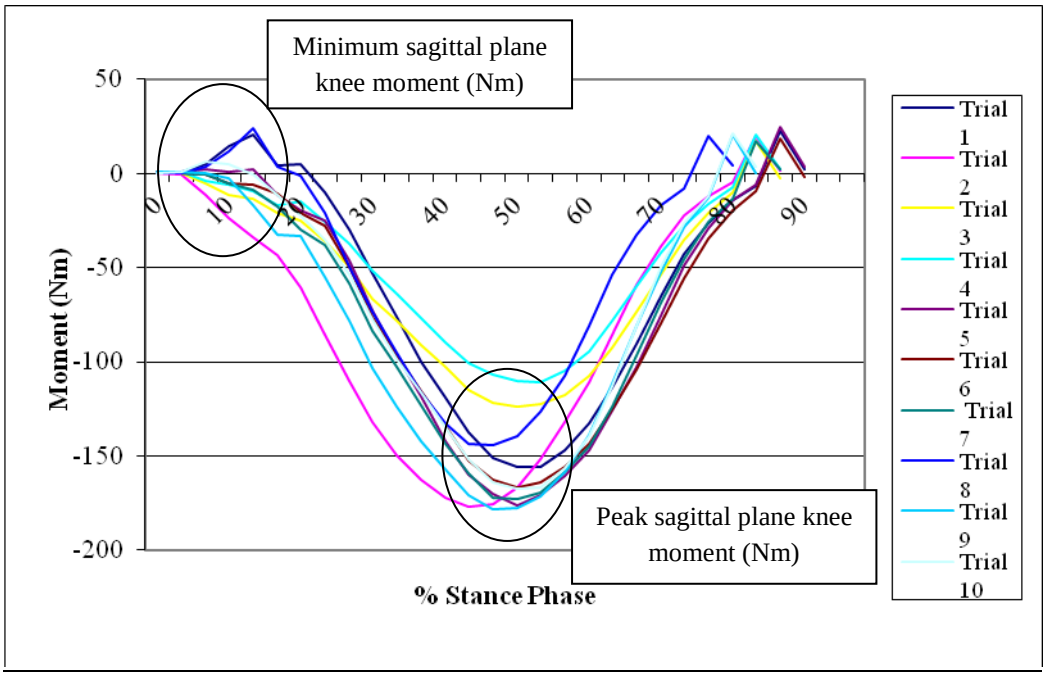


Figure 4.9. Sample moment graph illustrating the sagittal plane knee moment over the stance phase of gait (subject 10, mature).

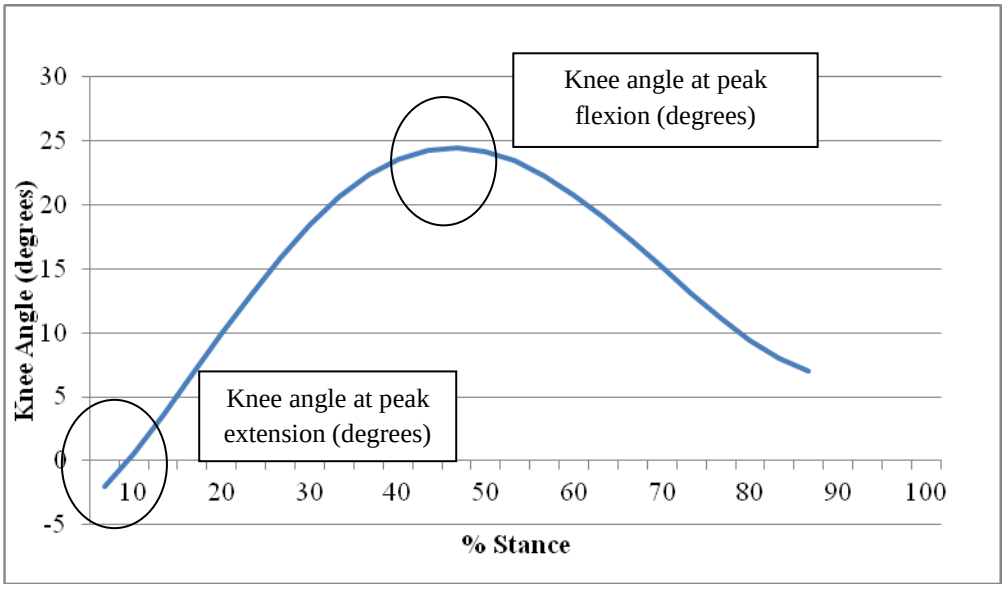


Figure 4.10. Sample graph illustrating the knee angle throughout the stance phase of running gait (mean data, subject 6 mature).

$$\Delta\theta = \text{knee angle at peak flexion} - \text{knee angle at peak extension}$$

Equation 4.6

$$\Delta M_k = \text{Maximum sagittal plane moment} - \text{minimum sagittal plane knee moment.}$$

Equation 4.7

$$\text{Knee joint stiffness} = \Delta\theta / \Delta M_k$$

Equation 4.8

Sagittal knee joint moment was calculated using an inverse dynamics technique, with inertia values provided by Dempster (1955). Change in knee angle was calculated instantaneously as the difference between the maximum angle at peak flexion (midstance) and the knee angle at heel strike (Equation 4.6). A similar technique was then employed to calculate the change in sagittal plane knee moment, with minimum moment at heel strike taken from maximum moment during midstance (Equation 4.7). Direction for the positive knee moment is illustrated in the diagram; moments acting in the anticlockwise direction are positive, while moments acting clockwise are negative (Figure 4.8).

4.3.5.3. Muscle strength assessment.

Quadriceps strength was assessed with the use of a Biodex Isokinetic Dynamometer (System 3, IPRS Mediquipe, Suffolk, UK). This assessment provided information on the strength of the quadriceps femoris muscles during single leg extension exercises, and has been shown to be reliable for knee flexion/extension measurements (Ly & Handelsman, 2002). Muscle

strength was assessed prior to running trials to reduce the possibility of fatigue influencing results. Figure 4.11 provides an illustration of muscle strength assessment.

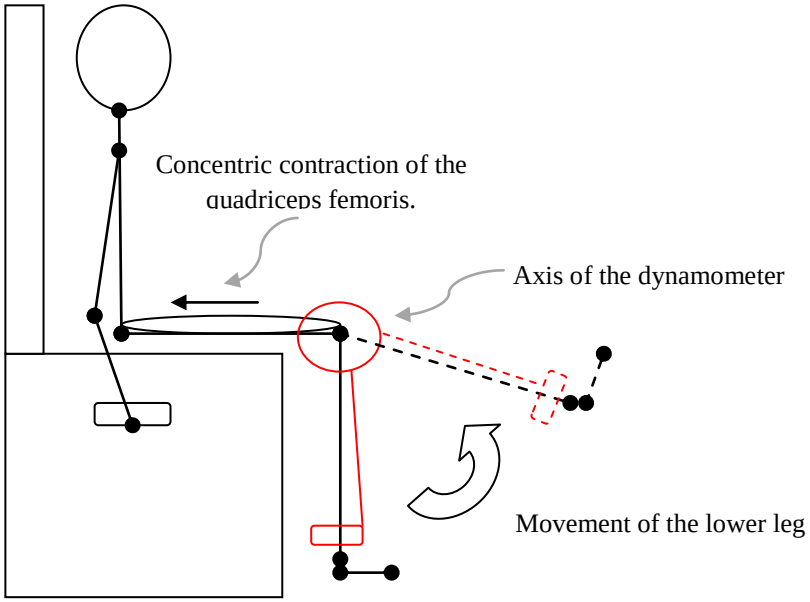


Figure 4.11. Illustration of a participant seated on the dynamometer, indicating knee extension movement.

Quadriceps strength was measured isometrically, with the participant seated on the dynamometer. Stabilisation was achieved with various straps, and handles affixed to the chair. One strap was placed around the waist and two around the tested leg; midway along the upper quadriceps, and around the ankle joint. The non-tested leg was strapped around the ankle to restrict any unwanted movement. In this assessment, only the right leg was tested. While seated, both the hip and the knee joints were positioned at approximately 90 degrees; the axis centre of knee flexion at the tibiofemoral joint was aligned by the eye with the axis of the dynamometer. The weight of the lower leg was measured with an inbuilt dynamometer facility for automatic gravity correction. The distal end of the shin attachment was placed at

approximately 5 cm above the lateral malleolus of the right leg, and adjusted based on comfort for the participant. After several practice attempts, followed by a five minute rest period, participants performed three maximum voluntary contractions of the quadriceps through extending the lower leg at the knee joint. Each contraction lasted approximately three seconds, with a minimum rest period of five seconds between each contraction. Maximum voluntary quadriceps force for each contraction was taken, and a mean for each participant was calculated. An independant sample t-test was employed to determine significant differences in maximum quadriceps strength between the mature and young females.

4.4. Results.

4.4.1. KOOS results.

The results from the KOOS (Table 4.1) demonstrate that both groups (young, 90.5; mature, 96.5) had minimal symptoms of osteoarthritis and knee injuries (KOOS, English Version LK 1.0). This was the case for both the left and the right knee. However it was observed that the Sport and Recreation, and Quality of Life subscales illustrated lowest scores (young, 93.8 and 94.1; mature 87.8 and 87.9).

4.4.2. Kinematic results.

Individual results are presented in Appendix E. Mean results from the kinematic assessment are presented in Table 4.3. Tables 4.4 a and b show a summary of the significance values for comparisons between mature and young females in both footwear conditions (a) and between the motion control and the neutral shoes among mature and young females (b).

Table 4.3. Kinematic results for both conditions among the mature and young female runners.

MATURE FEMALE RUNNERS		Component of variable measured during the stance phase of gait									
		Initial Angle (deg)		**Peak Angle (deg)**		**Occurrence time (sec)**		**ROM (deg)**		**Peak Angular Velocity (deg/s)**	
		Mature	**Young**	**Mature**	**Young**	**Mature**	**Young**	**Mature**	**Young**	**Mature**	**Young**
REARFOOT EVERSION	**N**	3.3 ± 5.2	3.0 ± 6.0	17.0 ± 5.6	12.06 ± 3.9*	0.12 ± 0.04	0.12 ± 0.05	11.9 ± 4.5	14.0 ± 4.7	308.8 ± 225.5	355.2 ± 216.8
	MC	1.97 ± 6.7	2.0 ± 6.5	10.91 ± 4.1 γ	8.8 ± 3.3*γ	0.12 ± 0.04	0.12 ± 0.06	9.4 ± 4.9	13.2 ± 7.3	307.2 ± 114.6	325.3 ± 89.2
ANKLE DORSIFLEXION	**N**	12.3 ± 13.1	18.4 ± 22.5	26.2 ± 6.5	33.9 ± 12.6	0.13 ± 0.04	0.1 ± 0.04	22.2 ± 11.9	18.0 ± 14.5	613.3 ± 308.3	656.3 ± 257.5
	MC	12.6 ± 15.7	18.9 ± 14.4	26.2 ± 3.6	34.3 ± 12.6	0.12 ± 0.02	0.09 ± 0.04	21.9 ± 12.4	16.5 ± 8.6	572.6 ± 204.6	504.2 ± 216.2
KNEE FLEXION	**N**	3.64 ± 2.5	1.74 ±1.92*	32.3 ± 11.4	28.9 ± 12.12	0.14 ± 0.07	0.14 ± 0.05	39.4 ± 19.8	40.7 ± 10.5	483.0 ± 149.6	497.9 ± 105.7
	MC	3.32 ± 2.6	1.45 ± 1.2*	32.2 ± 9.2	30.5 ± 12.65	0.15 ± 0.0	0.13 ± 0.04	36.8 ± 19.2	38.0 ± 8.9	463.8 ± 116.4	460.5 ± 80.4
KNEE ABDUCTION	**N**	15.2 ± 3.2	12.2 ± 5.3	23.1 ± 5.1	17.0 ± 5.96*	0.23 ± 0.02	0.23 ± 0.03	13.6 ± 6.9	19.5 ± 4.6	93.0 ± 42.5	99.6 ± 34.0
	MC	16.3 ± 4.0	11.2 ± 8.8	24.0 ± 6.5	17.1 ± 6.38*	0.23 ± 0.03	0.22 ± 0.04	15.4 ± 8.9	19.6 ± 7.8	85.8 ± 12.5	88.6 ± 21.8
KNEE INTERNAL ROTATION	**N**	11.2 ± 10.3	11.6 ± 5.5	-18.4 ± 3.7	-14.3 ± 2.1*	0.10 ± 0.03	0.11 ± 0.07	16.5 ± 17.6	9.3 ± 10.3	108.8 ± 55.0	99.2 ± 34.6
	MC	13.3 ± 7.9	13.4 ± 7.1	-15.1 ± 3.1γ	-12.6 ± 2.6*	0.11 ± 0.05	0.1 ± 0.07	17.1 ± 14.8	11.4 ± 12.2	87.9 ± 42.4	94.9 ± 16.5

Significance is highlighted within the tables with the symbols * and γ, denoting a significant difference between young and mature females (*) and motion control shoes and neutral condition (γ) at the $p < 0.05$ level.

Tables 4.4 a and b. Tables presenting p values from the variables showing statistically significant differences.

a)

	Mature : Young (Neutral)	Mature : Young (Motion Control)
Peak Rearfoot eversion	**P = 0. 03***	P = 0.34
Initial knee flexion	**P = 0.008**	**P = 0.007**
Knee abduction	**P = 0.04***	**P = 0.03***
Knee internal rotation	**P = 0.005***	**P = 0.06***

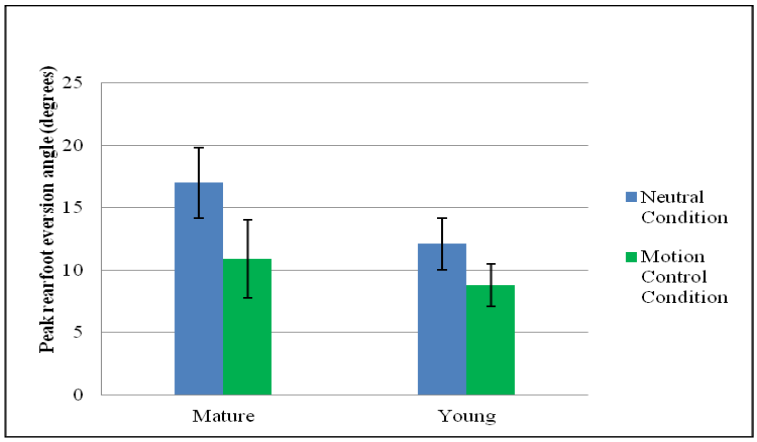
b)

	Motion Control: Neutral (Mature)	Motion Control: Neutral (Young)
Rearfoot eversion	**P = 0.03***	**P = 0.05***
Knee internal rotation	**P = 0.04***	P = 0.14

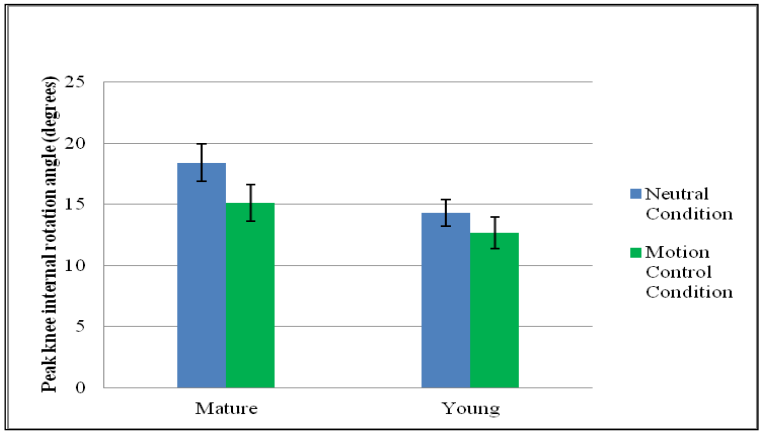
4.4.2.1. Significant kinematic results.

The figures below illustrate the differences in the peak rearfoot eversion (a), knee internal rotation (b), and knee abduction angles (c) between the two groups in both footwear conditions.

(a)



(b)



(c)

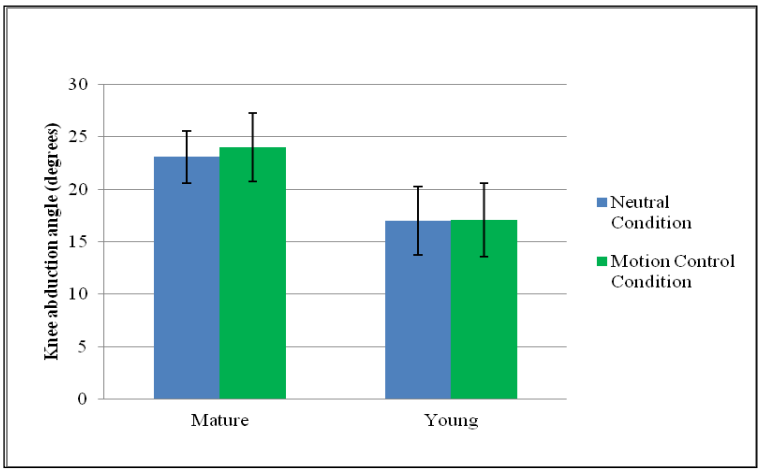


Figure 4.12. Bar graphs illustrating difference in rearfoot eversion (a), knee internal rotation (b) and knee abduction angle (c) between mature and young females in both footwear conditions.

4.4.2.1.1. Rearfoot eversion.

Eversion of the calcaneus occurred during the stance phase of gait, with peak rearfoot eversion occurring shortly after heel strike among both groups (0.12 ± 0.015 seconds). Results of the two way between-within ANOVA suggested that the peak rearfoot angle was significantly higher for the mature females compared to the young group in the neutral footwear condition ($p < 0.05$). Trials performed in the motion control condition however showed no significant difference between mature and young females ($p > 0.05$). Further analyses showed a significant difference between the motion control and the neutral condition among both groups ($p < 0.05$). Despite this, no significant difference was demonstrated between mature and young, or between footwear conditions in the initial angle, occurrence time of peak angle, range of motion or angular velocity of eversion ($p>0.05$). A sample angle time history throughout the stance phase of gait is displayed in Figure 4.13, showing single trial data for both a mature and young female in both footwear conditions.

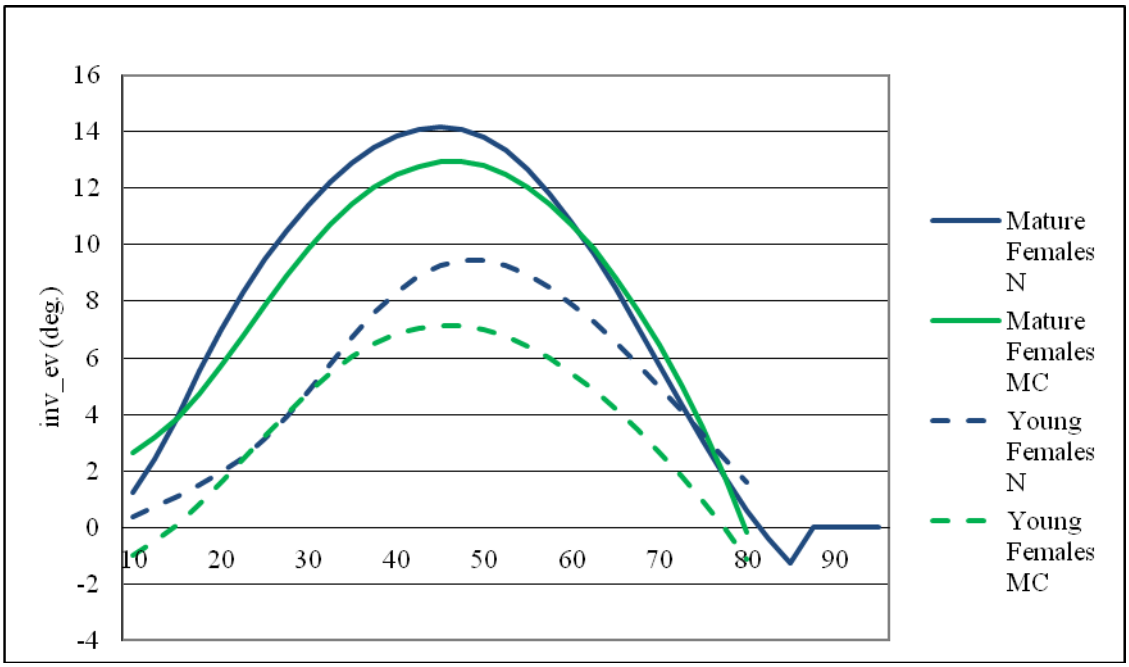


Figure 4.13 Rearfoot eversion graph illustrating single trial data from Subjects 6 (y) and 9 (m) in both footwear conditions.

4.4.2.1.2. Knee internal rotation.

Peak knee internal rotation was measured throughout stance, and peaked at an average of 0.1 ± 0.05 seconds. Peak knee internal rotation angles were significantly greater among mature compared to younger females, and were significantly reduced among the mature group with the use of motion control shoes ($p < 0.05$). However the motion control shoes did not significantly affect peak knee internal rotation values among the younger group ($p > 0.05$). No significant differences were observed in initial angle, occurrence time of peak knee internal rotation, maximum range of motion and angular velocity between groups or footwear condition.

4.4.2.1.3. Knee abduction angle.

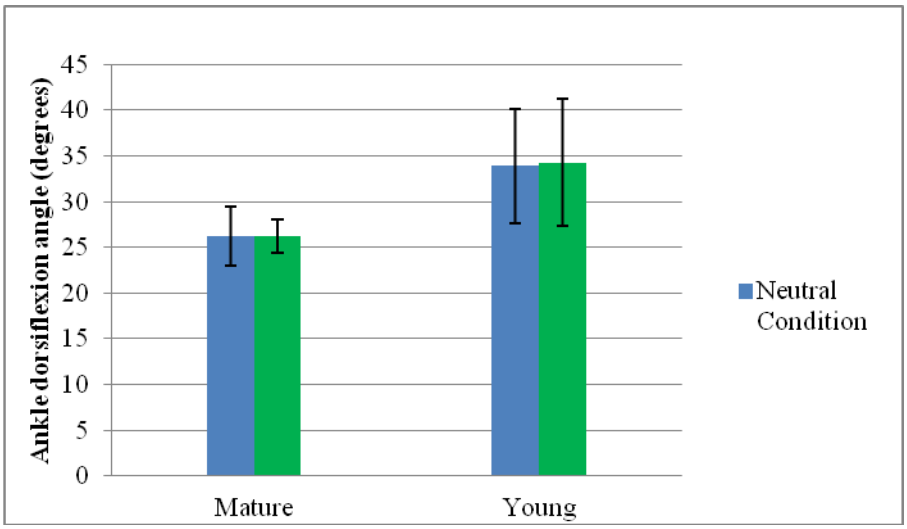
Peak knee abduction angle calculated throughout the stance phase of running gait was significantly greater for the mature group compared to the younger group in both footwear conditions ($p<0.05$). However no significant difference was seen in peak angle when comparing the trials performed in the neutral shoe with the motion control condition among either group ($p>0.05$). Similarly, no significant difference was seen in the initial angle, the occurrence time of peak angle, or the angular velocity of peak knee abduction ($p>0.05$).

4.4.2.2. Non-significant kinematic results.

Peak ankle dorsiflexion and peak knee flexion angles both elicited no significant differences between groups or with the different footwear conditions ($p > 0.05$). A significant between group difference was however seen in the initial knee angle ($p<0.05$). However no differences were demonstrated in initial angle of dorsiflexion. Similarly, no

significant difference was seen in occurrence time of peak angle, range of motion or angular velocities for either kinematic variable ($p > 0.05$). Bar charts highlighting the consistency between groups and trends with footwear conditions are highlighted for ankle dorsiflexion and knee flexion, in Figures 4.14a and b respectively.

(a)



(b)

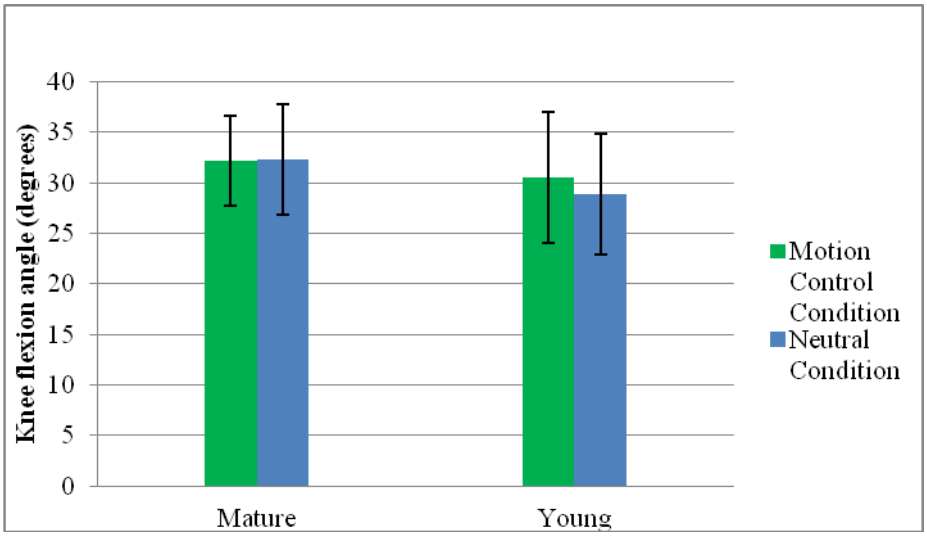
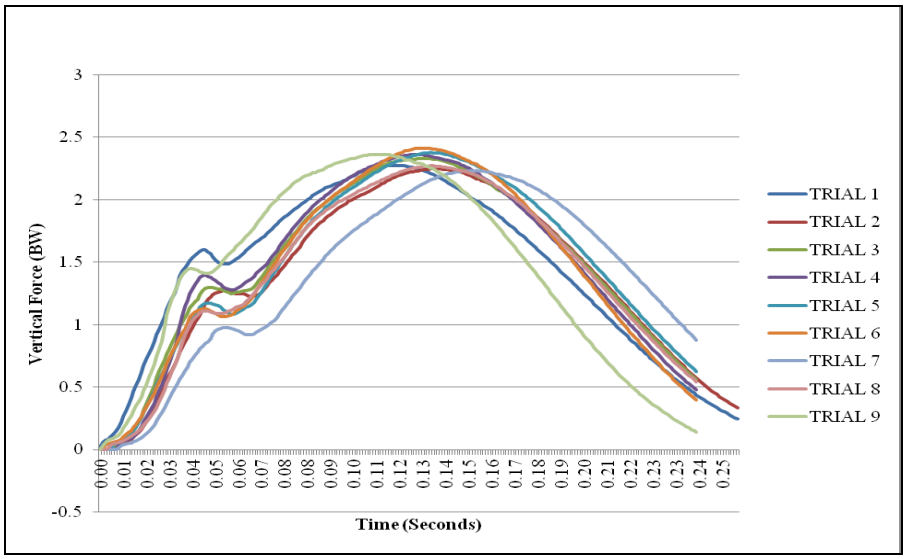


Figure 4.14. Graph illustrating mean data for ankle dorsiflexion (a) and knee flexion (b).

4.4.3. Kinetic results.

A sample vertical force time history for a mature female in both footwear conditions is presented in Figures 4.15 a and b, illustrating the impact force peak. Mean results for each participant are displayed in the appendix (Appendix E). Results from the statistical analysis for difference in the means are presented in Table 4.5. For the ground reaction force variables assessed, no significant effect was demonstrated for age, or footwear conditions among the two groups (p>0.05).

(a)



(b)

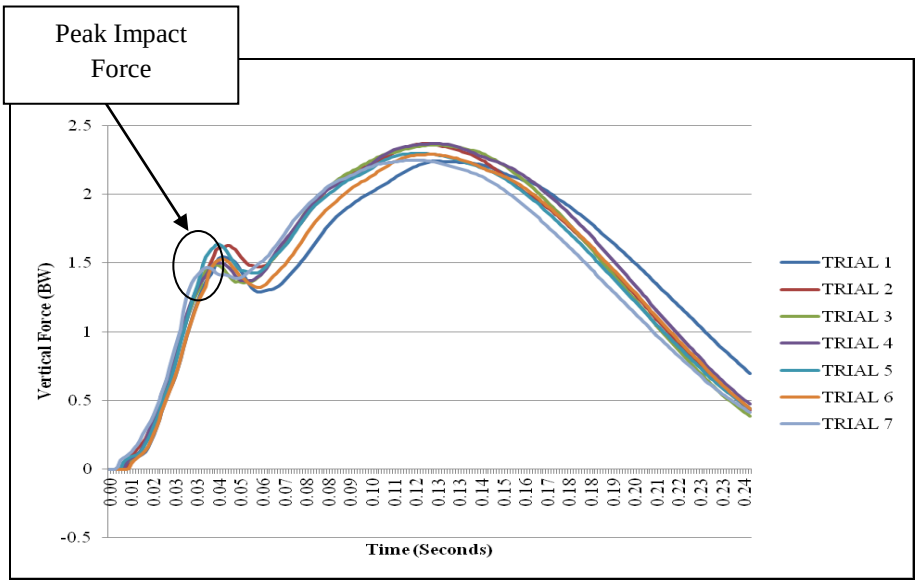


Figure 4.15 a and b. Sample ground reaction force-time history for participant 19 (m) in neutral (a) and motion control (b) footwear conditions.

Table 4.5. Statistical output from a two-way ANOVA for the effect of age and footwear conditions.

P-Values	**MOTION CONTROL Vs NEUTRAL**	
YOUNG	*Pk Fz*	0.27
	Occ Time PkFz	0.46
	Loading rate (LR)	0.58
	Occ Time LR	0.44
MATURE	*Pk Fz*	0.81
	Occ Time PkFz	0.80
	Loading rate (LR)	0.94
	Occ Time LR	0.58
	MATURE Vs YOUNG	
NEUTRAL	*Pk Fz*	0.23
	Occ Time PkFz	0.21
	Loading rate (LR)	0.84
	Occ Time LR	0.18
MOTION CONTROL	*Pk Fz*	0.49
	Occ Time PkFz	0.77
	Loading rate (LR)	0.68
	Occ Time LR	0.59

4.4.4. Peak knee external adductor moment and moment arm.

Peak knee external adductor moment was measured as the second peak on the moment-time history. A sample knee external adductor moment time graph highlighting the second peak is displayed in Figure 4.16. This graph illustrates a difference of 42.31 Nm between the two peaks; the greatest difference seen in any raw trial data between the mature and the young female runners in the neutral condition. Mean moment and moment arm data from the two groups for the two footwear conditions is presented in Table 4.6. Results illustrate a significant difference in peak knee external adductor moment between the mature and the young in both the motion control and the neutral footwear conditions ($p<0.05$). A similar significant difference was seen in the moment arm length ($p<0.05$). The effect of the motion control shoes was not significant in either group for the peak moment (young, $p = 0.83$, mature, $p = 0.81$) or the moment arm (young, $p = 0.36$, mature $p = 0.10$).

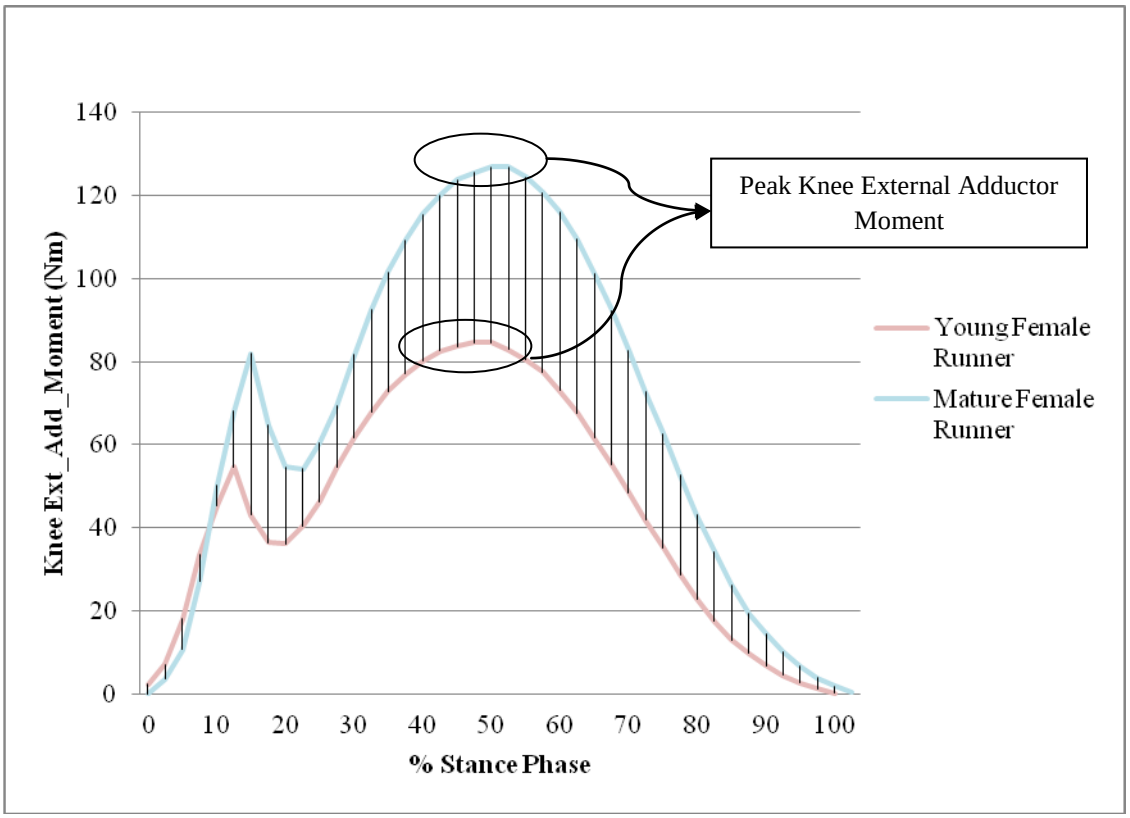


Figure 4.16. Sample knee external adductor moment for a mature and young female performing one running trial in the neutral footwear condition.

Table 4.6 Mean knee external adductor moment data, displaying peak values and moment arm length.

(a)

	NEUTRAL		**MOTION CONTROL**	
	Mature	**Young**	**Mature**	**Young**
Mean Peak Knee External Adductor Moment (Nm/kg)	1.45 (±0.17)	1.11 (0.12)	1.48 (±0.24)	1.13 (0.14)
Significance level (0.05)	**P = 0.0008***		**P = 0.002***	
Mean Knee External Adductor Moment Arm (cm)	3.0 (±0.05)	2.5 (±0.05)	3.2 (±0.03)	2.8 (±0.03)
Significance level (0.05)	**P = 0.037***		**P = 0.014***	

Figure 4.17 illustrates the association between the knee external adductor moment and the knee abduction angle produced by one mature female throughout the stance phase of gait. Mean data for both variables was used to represent all running trials performed in the neutral condition. Figure 4.18 illustrates the moment arm length throughout the stance phase of gait. As described previously, positive moment arm illustrates an adducted knee joint, whereby moment arm length calculated as a negative value illustrates an abducted (varus) knee joint position.

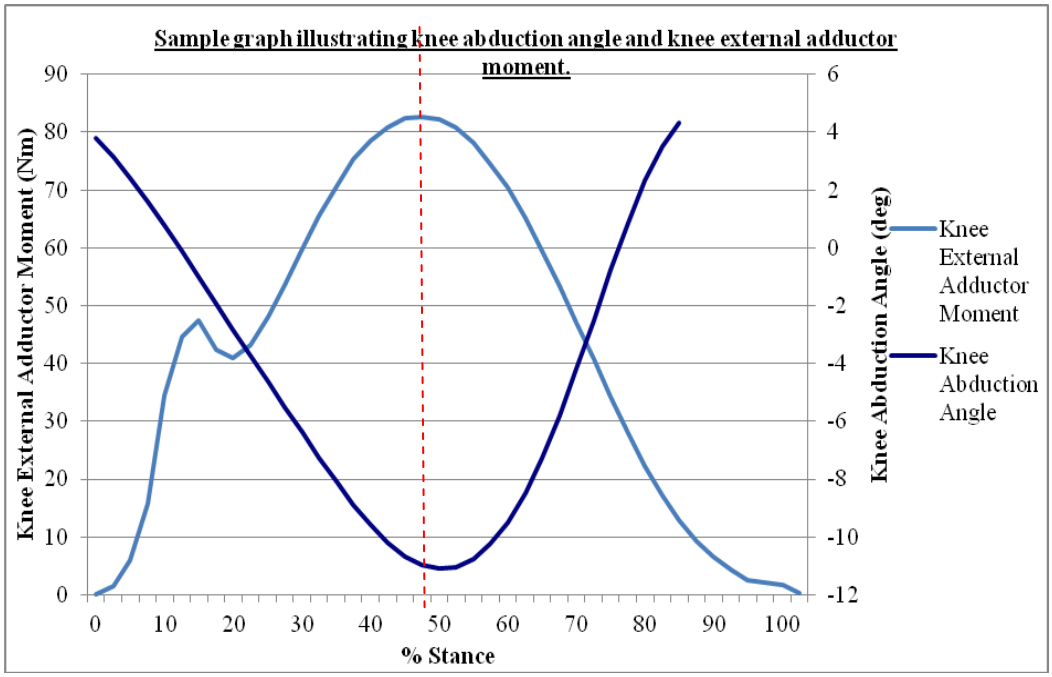


Figure 4.17. Sample graph illustrating sample knee abduction angle and sample knee external adductor moment trace produced by Subject 10 (young) in the neutral condition.

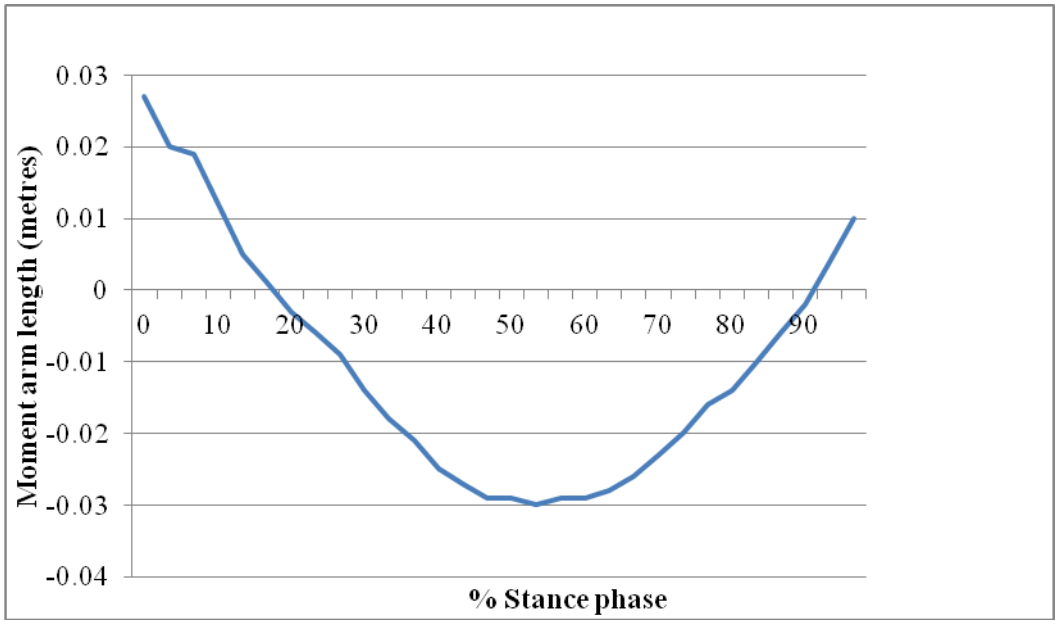


Figure 4.18. Moment arm length calculated throughout the stance phase of gait for Subject 10 in the neutral condition.

4.4.5. Knee joint stiffness.

Knee joint stiffness was assessed using the equation 4.8. Figure 4.19 illustrates a sample moment-angle graph for one mature participant performing running trials in the neutral footwear condition. Figure 4.20 illustrates the independent trace for the moment and the angle throughout stance, presented as a moment-angle graph. Both heel strike, toe off and peak knee joint flexion are highlighted on the graph. Table 4.7 displays mean results from the two groups, and the results of a two way ANOVA. These results indicate a significantly greater knee joint stiffness value for the mature compared to the young female runners in the neutral condition ($p<0.05$), and in the motion control shoe ($p<0.05$). When comparing between footwear conditions, differences between the motion control and the neutral condition were not significant for either group ($p>0.05$).

Table 4.7. Mean data for knee joint stiffness for mature and young participants in the neutral footwear condition.

Mean Knee Joint Stiffness (Nm/°/kg)	**NEUTRAL**		**MOTION CONTROL**		***Between condition significance (p<0.05) for Moment (M) and Stiffness (S)***
	Peak Sagittal Plane Knee Moment (Nm)	**Peak Knee Joint Stiffness (Nm/deg/kg)**	**Peak Sagittal Plane Knee Moment (Nm)**	**Peak Knee Joint Stiffness (Nm/deg/kg)**	
MATURE	154.56 (25.37)	0.21 (±0.077)	159.74 (±19.34)	0.23 (±0.06)	M: p = 0.39 S: p = 0.41
YOUNG	143.04 (29.52)	0.13 (±0.05)	148.35 (±24.093.)	0.15 (±0.03)	M: p = 0.28 S: p = 0.31
Between group significance (p<0.05)	p = 0.28	**p = 0.014***	p = 0.34	**p = 0.017***	

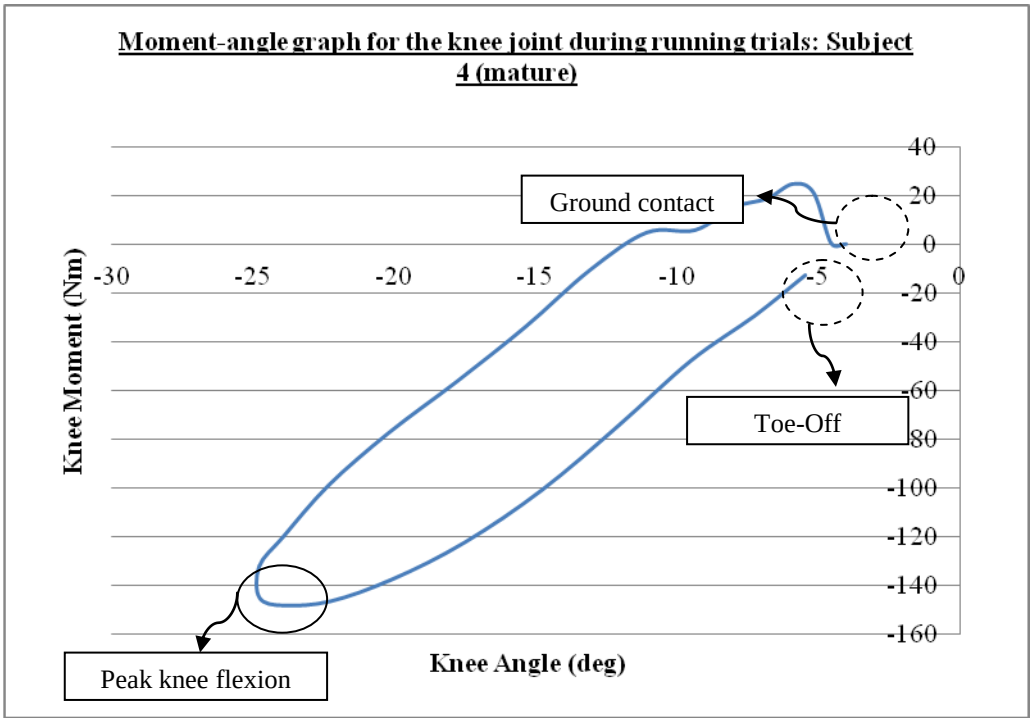


Figure 4.19. Sample moment-angle graph for the knee joint during running trials performed by a mature participant (4) in the neutral condition.

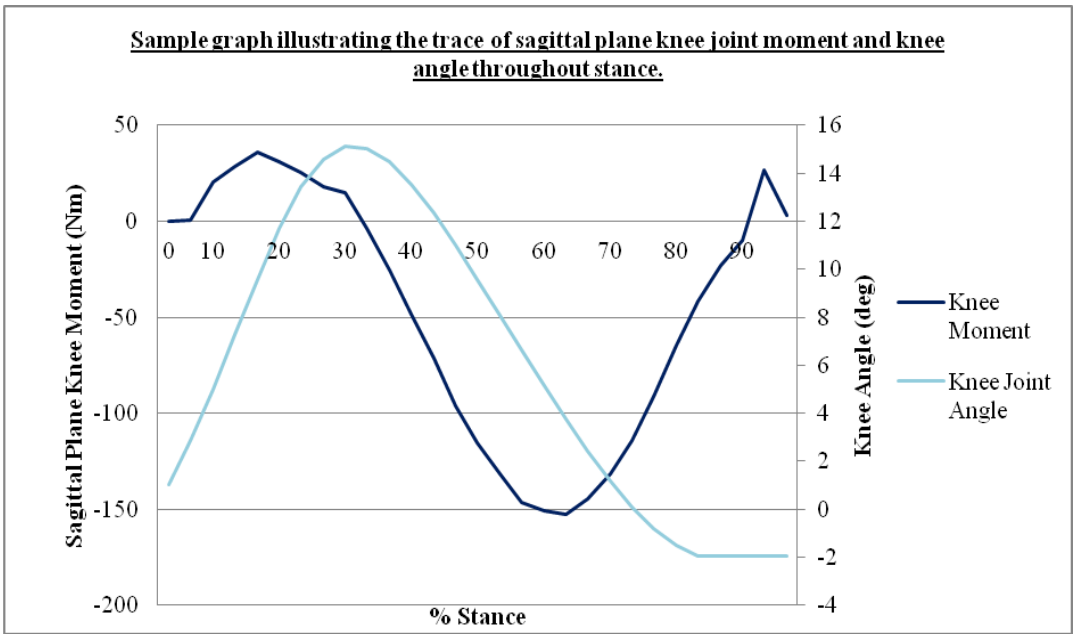


Figure 4.20. Sample trace of the sagittal plane knee moment and the knee angle produced by a mature participant (10) in the neutral condition.

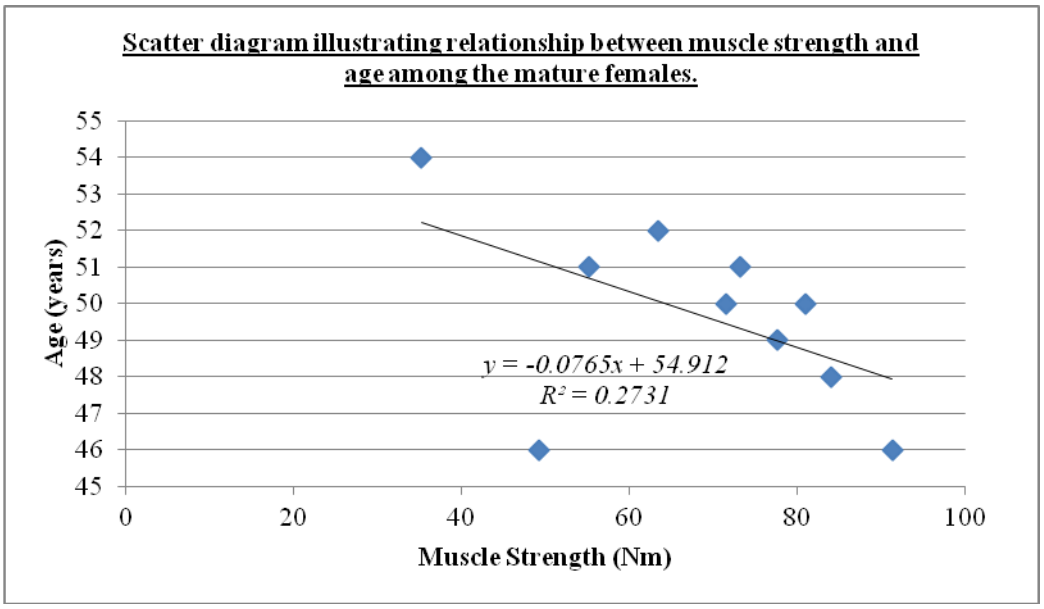
4.4.6. Quadriceps Muscle strength.

Table 4.8 displays both mean results for quadriceps femoris muscle strength for each participant, and group means for the mature and the young female runners. Although the young group produced a mean muscle strength of 10.66 Nm greater than the mature group, the use of an independent sample t-test showed this difference to be non-significant ($p>0.05$). Figures 4.21 a and b illustrate the association between strength and age (a) and body mass (b) among the mature group. Results from a regression analysis for strength and age showed the relationship to be positive, but not significant ($r = 0.52$, $p = 0.12$), as was that between strength and body mass ($r = 0.41$, $p = 0.23$).

Table 4.8. Mean torque generated by the quadriceps for each participant, averaged from three trials.

	Mature Participant		**Young** Participant	
Mean maximum quadriceps strength, averaged from 3 trials (Nm).	1	77.5 (± 3.4)	1	70.2 (±1.2)
	2	80.9 (±1.6)	2	90.1 (±2.8)
	3	35.2 (±2.6)	3	88.3 (±0.9)
	4	55.1 (±2.0)	4	72.7 (±2.0)
	5	91.3 (±1.5)	5	81.7 (±1.7)
	6	49.2 (±1.6)	6	100.0 (±1.3)
	7	63.4 (±0.8)	7	61.4 (±1.1)
	8	71.4 (±2.5)	8	59.3 (±0.8)
	9	73.1 (±2.1)	9	77.7 (±1.5)
	10	84.0 (±0.2)	10	61.4 (±1.0)
			11	95.9 (0.5)
			12	86.6 (±1.8)
Mean	**68.12**		**78.78**	
Standard Deviation	**17.36**		**13.91**	
P-Value	**P = 0.17**			

(a)



(b)

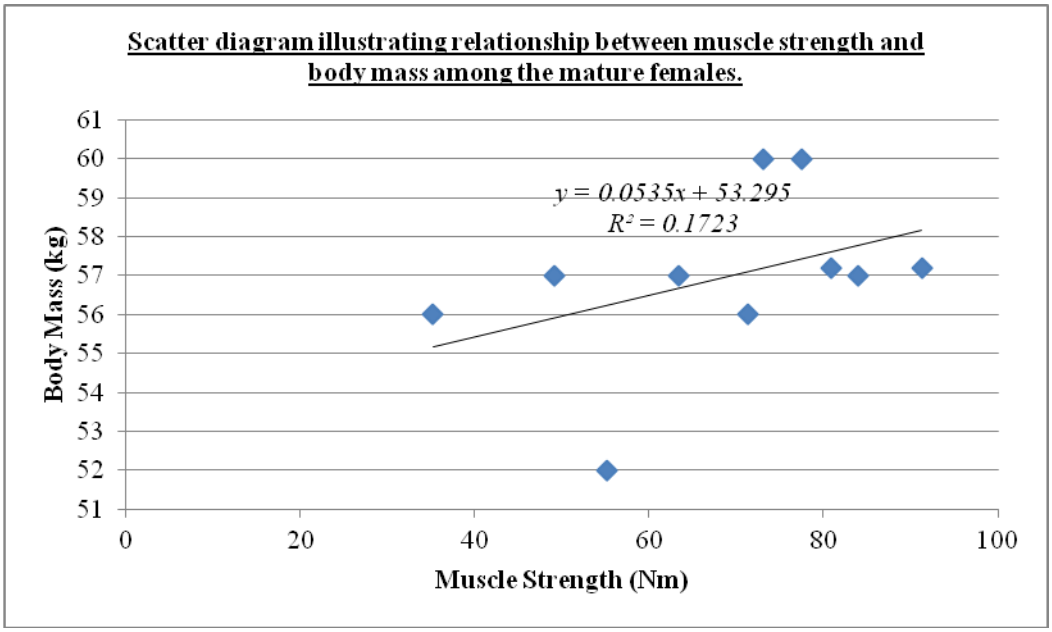


Figure 4.21a and b. Scatter diagram illustrating relationship between muscle strength and age (a) and muscle strength and body mass (b) for the mature females.

4.4.7. Summary of results and overall findings.

Compared with young females, the mature females exhibited:

- Significantly greater peak ankle eversion angle (neutral).
- Significantly greater peak knee internal rotation angle (neutral and motion control).
- Significantly greater peak knee abduction angle (neutral and motion control).
- Significantly greater initial knee angle (neutral).
- Significantly greater peak knee external adductor moment (neutral and motion control).
- Significantly greater peak knee external adductor moment arm (neutral and motion control).
- Significantly greater knee joint stiffness (neutral and motion control).

Compared with the neutral shoe, the motion control shoe resulted in:

- A significant decrease in ankle eversion angle (mature and young).
- A significant decrease in peak knee internal rotation angle (mature).
- No significant change in peak knee external adductor moment or moment arm length.

4.5. Discussion.

The posture of the foot and ankle complex has long been considered to contribute to the development of lower limb musculo-skeletal injuries and conditions, due to the effect on mechanical alignment and dynamic function during gait (Levinger, Menz, Menz, Fotoohabadi, Feller, Bartlett, Bergman, 2010). As such, attention has been given to the differing influence of footwear as both an injury treatment and prevention method. This study aimed to compare the running gait of young and mature females while wearing a controlled neutral shoe, and to investigate the influence of a motion control shoe on running biomechanics. Twenty two females, ten mature (40-60 years) and twelve younger (18-25 years) performed running trials in two footwear conditions; a neutral and a motion control running trainer. Alongside the gait analysis assessment, all females were assessed for maximum strength of the quadriceps femoris muscle.

It was hypothesised that the running trials performed in the neutral shoe would result in a difference in the gait of mature compared to younger females; specifically, higher peak values of rearfoot eversion, knee internal rotation, knee external adductor moment and knee joint stiffness. Secondly, it was hypothesised that the motion control shoe would reduce the biomechanical variables of rearfoot eversion and knee internal rotation among all females, with greater modifications for the mature group. Thirdly, it was hypothesised that the medial support in the motion control shoe would increase the knee abduction and the moment arm length, producing an overall rise in the peak knee external adductor moment in both groups. Finally, in terms of physiological differences, it was hypothesised that the muscle strength of the mature group would be significantly lower than that of the younger group.

4.5.1 Hypothesis One: A controlled neutral shoe.

In support of hypothesis one, when running in the controlled neutral shoe, rearfoot eversion, knee internal rotation, knee external adductor moment, and knee joint stiffness were significantly higher among mature (40-60 years) compared to young females (18-25 years). Although this theory was suggested in the previous chapter, the results from this current investigation support the notion that, after controlling for the effect of footwear, the biomechanics of running gait were significantly different among a group of mature females when compared to a younger group.

4.5.1.1. Kinematic differences in the running gait of mature and young females in a controlled neutral footwear condition.

Rearfoot eversion is a natural mechanism involved in shock attenuation during the stance phase of walking and running gait (Nordin & Frankel, 1989). However, it has been extensively reported that excessive magnitudes of rearfoot eversion lead to increased risk of developing lower limb conditions (Clarke et al., 1983; Hintermann & Nigg, 1998; O'Connor et al., 2004).

The increased rearfoot eversion demonstrated by mature females leads to a suggestion that increasing age results in altered biomechanics. An early study by Hutton and colleagues suggested that advancing age causes a "sag" in the longitudinal arch of the foot, increasing the proportion that is in contact with the ground, and producing a pronated position at the subtalar joint (Hutton & Dhanendran, 1979). This was later confirmed by the theory that changes in foot structure and biomechanics occur with advancing age, and Staheli and colleagues indicated a trend towards flatter feet in people over the age of 30 years (Staheli, Chew & Corbett, 1987). Similarly, with the use of an arch index assessment, Scott and colleagues showed mature participants to present with higher levels of pronation compared to a younger group (Scott, Menz & Newcombe, 2007). Although the results from this latter study (Scott et al., 2007) were shown during both standing and walking gait, the results among the mature females appear to confirm

the suggestions that changes occur with age, with significantly greater values of rearfoot eversion compared to a younger group in a neutral footwear condition.

At the foot-ankle complex, excessive rearfoot eversion and subtalar joint pronation places increased pressure on the plantar fascia during weight-bearing; a common risk factor for the development of plantar fasciitis (Irving, Cook, Young, & Menz, 2007). The high magnitude of rearfoot eversion seen among the mature females therefore supports suggestions in the literature that increasing age is a common risk factor for the development of plantar fasciitis among runners (Scher, Belmont, & Owens., 2007). Therefore it could be suggested that the link between age and this condition is due to tendency for mature participants to run with an increased magnitude of rearfoot eversion. Increased rearfoot eversion and associated pressure on the plantar fascia during the stance phase of running, has been suggested to alter the biomechanics of the adjacent limbs and reduce the efficiency of force absorption (Kilber, Goldberg, Chandler, 1991). Excessive rearfoot eversion may therefore also lead to adverse effects at the proximal joints; specifically the knee joint.

According to Laughton and colleagues, excessive rearfoot eversion leads to increased strain on the soft tissues around the knee joint, and an increased risk of developing specific conditions associated with overloading (Laughton, McClay-Davis, & Hamill, 2003). It has been shown that increased eversion angle of the rearfoot and accompanied increased loading under the medial forefoot is significantly related to the development of exercise related lower leg pain (Willems et al., 2006). In addition, Reilly and colleagues illustrated a significantly greater frontal plane calcaneal angle, indicative of rearfoot eversion, among a group of patients presenting medial knee osteoarthritis (Reilly, Barker, Shamley, & Sandall, 2006). However there is no mention of the certainty that the rearfoot eversion angle was not a consequence of the altered knee biomechanics caused by the osteoarthritis, as opposed to a causative factor itself. The link between medial compartment knee osteoarthritis and higher values of rearfoot eversion was however also considered by Riegger and Krugh (1996) who suggested that people with higher levels of knee osteoarthritis and associated knee external adductor moments, also display higher values of subtalar pronation, to enable the foot to plantigrade (with the heel, metacarpals, carpals and phalanges all in contact with the ground) during the stance phase of gait. However, as these female runners were free from symptoms of osteoarthritis or injuries at

the knee joint, the possibility of subtalar joint pronation occurring as a result of biomechanical changes due to injury appears minimal. Instead, the high subtalar pronation appears more likely a risk factor for this population.

Although the mature females demonstrated a significantly greater peak eversion angle, no difference was seen in the occurrence time of the angle, or the velocity with which the angle reached its peak. This is contrary to previous research suggesting that an increase in peak angle is associated with a faster occurrence time, and it is the latter that is also associated with increased stress at the knee joint (Hamill, Emmerik, & Heiderscheit, 1999). Despite this, it is suggested that it is the peak angle of rearfoot eversion angle presented by the mature females that may contribute to the high injury rates documented for this population (Taunton et al., 2002; Hintermann & Nigg, 1998).

Due to the coupling between the subtalar joint and the tibia, and the inclination of the subtalar axis, rearfoot eversion is coupled to internal rotation of the tibia and knee joint (Nigg et al., 1993). This relationship has been supported in this study, with the significantly higher peak knee internal rotation angle presented by the mature females in the current study ($p<0.05$) likely related to the high rear foot eversion. Although the occurrence time of peak knee internal rotation angle was not different between the two groups, in the neutral condition, both rearfoot eversion angle and knee internal rotation peaked at an average of 0.11 seconds for both groups.

Knee internal rotation occurs during the early stages of the stance phase of gait (loading phase). This movement is dominated by the motion of the tibia rotating about the femur, and as such is commonly linked with the position of the foot and ankle on the ground (Richards, 2008). Knee internal rotation is partly controlled by the posterolateral structures of the knee joint, and as such is commonly linked with the increased risk of damage to the anterior cruciate ligament (ACL). Although this current study with the mature females did not directly consider ACL injuries, the association between this condition and knee internal rotation is discussed. It has commonly been suggested that in an ACL deficient knee joint, knee internal rotation is higher, with increases of 20% reported in the literature (Andriacchi & Dyrby, 2005). Similarly, it has also been reported that the strength of the ACL ligament structure and its insertion diminishes with advancing age (McKeon, Bono, & Richmond, 2009). As such, it is possible that reduced

flexibility and strength of the ligaments within the knee joint could be a contributing factor to the high level of knee internal rotation demonstrated by the mature females.

This high knee rotation for mature females is likely contributing to the higher injury rates that have been documented for this population (Taunton et al., 2002; Rose et al., 2011). Excessive pronation of the subtalar joint, and internal rotation of the knee, forces the patella out of the patellar grove of the femur to ride over the lateral aspect of the groove and irritate the lateral patella cartilage (Ramig, Shadle, & Watkins, 1980). The magnitude of rearfoot eversion and the transferral into tibial internal rotation, and internal rotation of the knee joint has therefore been related to overloading and stress at the lateral knee (Eslami et al., 2007). According to Andriacchi and Dyrby (2004), an altered rotational position of the knee joint could produce changes in the tibiofemoral contact area during gait, leading to degenerative changes in the meniscus and articular cartilage. Similarly, during gait, high values of knee internal rotation interfere with normal knee function as the patella is shifted laterally, in turn increasing the vertical pull of the vastas medialis, increasing compressive forces on the knee joint (Olson, 2007). Additionally, excessive internal rotation of the knee joint causes decreased patellofemoral contact area, in turn increasing the pressure in specific areas, increasing wear on the lateral condyles of the knee joint (Norris, 2004).

Although no significant difference was seen in the peak knee flexion angle between the two group, the initial knee angle was significantly greater among the mature compared to younger female runners ($p<0.05$). This suggests that at ground contact, the mature females had significantly more flexion angle at the knee joint, in contrast to the younger group who tended to strike with a straighter leg. During running gait, in preparation for heel strike, the hamstring muscles contract to control for knee hyperextension. Among the mature female runners, the increased knee flexion angle at heel strike could therefore be considered a possible consequence of tightness of this muscle group. According to Akinpelu, Bakare and Adegoke (2005), hamstring tightness significantly increases with age up to 49 years, suggesting that the hamstrings of the mature group will naturally be tighter than those of the younger group of females.

It has been shown that an increase in knee angle (flexion) at heel strike improves the impact attenuation during the early stance phase of running gait (McMahon, Valiant & Frederick, 1987). However, other researchers have suggested that an increase in knee

angle at ground contact has varying effects on impact attenuation above and below the knee joint; with impact attenuation of the shank shown to increase, and of the thigh, shown to improve (Hamill, Derrick & Holt, 1995; Derrick, Hamill & Caldwell, 1998). The consequence of knee angle at initial contact has been subject to debate, and will be discussed in greater detail in its relation to knee joint stiffness in a later section.

A significant difference was also observed in the magnitude of knee abduction angle, where mature females ran with a significantly greater peak knee abduction angle ($p<0.05$). This occurred during late stance among both groups. The increase in knee abduction among the mature females is of particular interest due to its association with the knee external adductor moment. Although it was not directly hypothesised that the mature group would display greater knee abduction angles, as the adductor moment was predicted to be greater, the position of the knee was expected to be in a more abducted (varus) position. An increase in knee abduction causes an increase in the moment arm length; the distance between the knee joint centre and the line of frontal plane ground reaction force, therefore increasing the knee external adductor moment. This therefore indirectly links higher values of knee abduction to an increased risk of medial compartment knee osteoarthritis among this mature group of runners; a notion supported in the following review of relevant research.

An abducted or varus knee during running causes medial displacement of the resultant force acting through the knee joint (Maquet, 1976). This in turn has been shown to place specific areas of the knee under increased loads, with a varus knee causing a fourfold increase in the risk of medial tibiofemoral osteoarthritis, determined by the presence of tibiofemoral osteophytes (Sharma et al., 2001). Therefore, among the mature females, the high knee abduction values are considered a potential risk factor for the development of conditions at the knee joint including medial compartment osteoarthritis.

In support of this Chang et al (2011) suggested varus tibiofemoral malalignment in the frontal plane can be regarded as a typical marker of osteoarthritis disease severity, and a risk factor for unicompartmental progression of osteoarthritis of the medial knee (Chang et al, 2011; Janakiramanan et al., 2008). Chang and coworkers also related dynamic varus thrust, implying a lateral movement of the knee joint, during ambulation to be an important factor in the progression of medial compartment knee osteoarthritis (Chang et al., 2011). Additionally, Cahue, Dunlop, Hayes, Song, Torres, & Sharma (2004)

demonstrated a varus alignment of the knee to significantly increase the risk of osteoarthritis to the medial patellofemoral compartment, due to a varus shift in the collinear alignment of the hip, knee and ankle. This varus alignment in comparison to a "normally" aligned knee is illustrated in Figure 4.22.

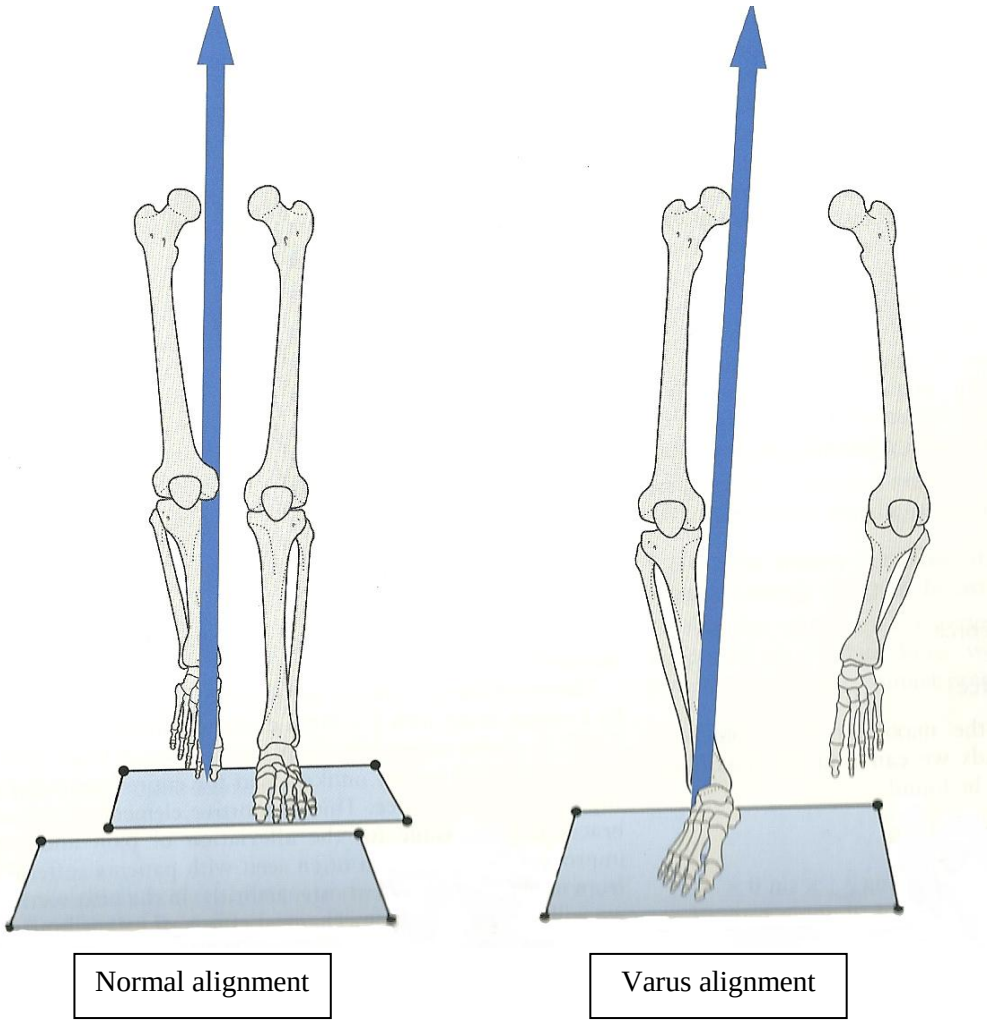


Figure 4.22. Comparison of the normal and varus aligned knee during gait.

Source: Richards, 2008.

A malaligned knee, with varus alignment has also been shown to increase knee valgus-varus laxity of the joint (Chang, Hayes, Dunlop, Hurwitz, Song & Cahue, 2004). Varus malalignment causes repetitive stretching of the collateral ligament, which increases capsule ligamentous laxity (Chang et al., 2004). Varus-valgus laxity at the knee joint has been related to development and progression of knee osteoarthritis, and has been shown to precede osteoarthritis development, and may predispose individuals to the condition (Sharma, Lou, Felson, Dunlop, Kirwin-Mellis, & Hayes, 1999). Similarly, laxity at the knee has been shown to increase with age, and is more common among women than men. In an investigation of 450 participants, radiographic signs of osteoarthritis were found in 18.4% of men and 26% of women. Interestingly, among women, of those with radiographic osteoarthritis, only 32.5% reported pain. Additionally, the incidence of valgus-varus laxity associated with radiographic osteoarthritis was significantly higher for women than men ($p<0.00001$), with varus laxity demonstrated in 48.5% of women and only 13% of men. The association of knee varus with medial compartment knee joint osteoarthritis is further discussed in a later section, in relation to the moment arm of the knee external adductor moment.

4.5.1.2. Kinetic differences in running biomechanics between mature and young females.

When running in the neutral shoe, as hypothesised a significant differences was observed in the knee external adductor moment between the mature and the young females. These results lie in accordance with those from the previous study; however this difference was observed when participants were wearing neutral shoes. This suggests that the differences in biomechanical joint moments are inherent to the mature group, as footwear has previously been shown to have an effect on knee external adductor moments. According to Yasuda and Sasaki (1987), an increase in the lateral support within running shoes may significantly reduce knee moments and medial compartment load. Therefore with no differing effect of footwear between the two groups, the significant difference in knee external adductor moment is considered a possible change inherent with increasing age among females.

The knee external adductor moment is an important biomechanical marker of joint pathology (Foroughi et al., 2009). Large knee external adductor moments are related to knee varus movements as described above, in turn increasing the loads at the medial compartment of the knee joint (Norris, 2004). This therefore lies in accordance with the kinematic results, indicating significantly greater knee abduction movements among the mature compared to the young females. The association between these two variables is shown in a sample graph (Figure 4.17), illustrating the two peaks to occur at a similar time in stance. This relationship is further supported by the results from the moment arm data. The higher moment arm produced by the mature group, suggests that the midpoint of the knee was located significantly further from the line of ground reaction force, when compared to the younger group. Furthermore, as all the moment arm values calculated were negative, this illustrates that the knee joint centre was positioned laterally to the line of ground reaction force, and the knee was in an abducted position.

As the results from the KOOS survey categorised all females as free from symptoms and signs of osteoarthritis, the high knee external adductor moments produced by this group suggest a high risk for the development of the condition. As discussed, the knee external adductor moment acts at the knee joint in response to an abducted knee position (Levinger et al., 2010). According to Miyazaki et al (2002) the knee external adductor moment is a strong predictor of medial compartment load and osteoarthritis, suggesting a possible reason for the high incidence and risk of this condition among this population. Furthermore, Sharma et al (1999) significantly associated knee external adductor moments with narrowing of the medial tibiofemoral compartment, suggesting increased magnitude of intrinsic compressive load. The significant difference between the mature and the young females in the magnitude of this moment suggest a possible association between age and altered joint torque. These results could therefore lie in accordance with the theory that an increase in age is a risk factor for knee joint osteoarthritis. Although the literature clearly categorises age as a risk factor for osteoarthritis development, little research has occurred to investigate the role of biomechanical changes with age. Therefore it is suggested that, with footwear controlled for, the mature females run with increased knee abduction and external adductor moments, which could increase susceptibility to conditions such as medial compartment knee osteoarthritis.

4.5.1.3. Biomechanical similarities between mature and young female runners.

In contrast to the significant changes in rearfoot eversion, knee internal rotation, knee abduction, and knee external adductor moment, several biomechanical variables appeared to show no difference for the two age groups. As illustrated in Table 4.3, no significant differences were seen in the peak ankle dorsiflexion or peak knee flexion angles between the mature and the young female runners ($p>0.05$). Both motions during gait occur in the sagittal plane, and are therefore predominantly controlled by the antagonistic muscles around the femur and those around the shank (Norris, 2004). Ankle dorsiflexion occurs when the talus glides posteriorly on the tibia, and forms a stable, closed-pack position during midstance (Malliaras, Cook & Kent, 2006).

Average ankle dorsiflexion angle ranges between 20-30°; runners presenting values below this range have been suggested to be at an increased risk of damage to the ankle and knee joint (Norris, 2004). Reduced dorsiflexion during running prevents the ankle from reaching its stable closed pack position at midstance, causing instability of the joint, and increased vulnerability to hypersupination; an additional risk factor for injuries (Lindsay et al., 2008). Similarly, Pope, Herbert and Kirwan (1998) investigated running gait of 1093 male army recruits over 12 weeks of intensive training, and recorded 48 injuries to the lower limb. A significant correlation was then demonstrated between reduced ankle dorsiflexion range of motion and injury occurrence.

Among the mature females, running trials in the neutral conditions elicited an average peak ankle dorsiflexion of 26.2 ± 6.5 degrees, compared to 33.9 ± 12.6 degrees shown by the younger group. Therefore, since both values are within or above the range suggested by the literature it is suggested that neither group is showing signs of increased injury risk associated with sagittal plane motion at the ankle joint (Brukner & Khan, 2006).

In addition to ankle dorsiflexion, peak knee joint flexion was also similar between the two groups. Reduced peak knee joint flexion angle has been associated with increased tightness of the hamstring and glutei muscles and subsequent risk of associated injury to both (Norris, 2004). Additionally, it has been proposed that decreased knee flexion throughout the stance phase of gait elicits anterior tibial shear loads that are in high enough magnitudes as to damage the anterior cruciate ligament of the knee joint (Myer,

Ford, Khoury, Succop & Hewett, 2011). Consideration of this lack of difference in the peak knee flexion is further discussed in relation to strength of the quadriceps muscle and knee joint stiffness during running.

Although an association between ankle dorsiflexion and knee flexion exists, this is predominantly seen at the end of swing phase (DeLisa, Gans & Walsh, 2005). However, the two motions do occur concurrently during the initial stages of running gait; as the ankle dorsiflexors and the knee extensors are simultaneously activated, termed the loading response (Kirtely, 2006). This therefore supports the results from the mature and young females, as no differences were seen in either variable between the two groups.

When directly comparing results from the two groups of females, the lack of change in range of motion at the knee and ankle joint tends to lie in opposition to the theory that decreased joint range of motion is a common factor associated with ageing (Messier et al., 1991). The conclusion based on the results from the females suggests that a change in range of motion does not occur from the approximate ages of 20 to 50 years. These results do however support those of Roach and Miles (1991) who conducted a large study into the effect of aging. Here it was suggested that differences in range of motion in people aged between 25 and 74 years were minimal (3-5 degrees) and any change in range of motion in people before the latter age should be considered abnormal (Roach and Miles, 1991).

Similar to the kinematic variables of knee joint flexion and ankle dorsiflexion, the loading rates exhibited by the two groups were not significantly different in the neutral footwear condition. Each participant within this study was categorised as a heel striker, indicating that the initial contact with the ground at the end of swing phase occurs with the heel. The force attained here is the impact peak, and occurs within the first 10% of stance (Hreljac, 2004). Depending on speed and kinematic position of the ankle and foot complex, during running the impact force varies from 1.5 to 5 BWs. The rate of loading implies the speed with which this force is applied and transmitted to the lower limb, and has been consistently linked with overuse injury prevalence. The lack of difference in loading rates among the mature females compared to the younger group is surprising given the significant difference shown in the previous study.

Common theories have previously associated high loading rates with running-related overuse conditions; eliciting a direct risk to damage of articular cartilage and subchondral

bone of the knee joint (Syed & Davis, 2000). Nigg and colleagues recruited a group of both control and injured participants, and found the latter to have significantly higher rates of loading than the former group (Nigg et al., 1981). Similarly, Hreljac (2000) investigated a group of female runners with a history of various overuse conditions, and showed them to present significantly greater loading rates than a group of females that had not previously attained an overuse condition. In an additional study on female runners, Ferber et al (2002) reported that a history of stress fractures was positively associated with higher loading rates during running, although this investigation only incorporated injuries to the tibia as opposed to the knee joint.

Although all the mentioned studies appear to link high loading rates to injury risk, it is noted that the link is often retrospective, with previous or current injury status being correlated with high loading rates. This therefore questions the categorisation of loading rates as a lone risk factor for the development of injuries. This latter thought is supported by the results of one prospective study, which showed that subjects with greater loading rates during gait in fact sustained significantly fewer running related injuries than those with higher magnitudes of the kinetic variable (Nigg et al., 1997). Similarly, results from one study among post-menopausal women who were exposed to a two-year long programme of high impact exercise showed a 4 % increase in bone mineral density, compared to a decline among a non-impact group (Heinonen, Kannus & Sievanen., 1996). These findings therefore lie in contrast to previous suggestions that impact loading leads to injury.

It was initially hypothesised that the mature group would display significantly greater loading rates compared to the younger group, as the former are suggested to be more prone to injury occurrence. Although in a high risk category, the lack of higher loading rates among the mature group could be due to the fact that no participant had sustained a previous injury while running. Similarly, although mature females appear to be at a greater risk of injury and debilitating conditions due to certain biomechanical features, the results from this study alongside those in the literature question the importance of these loading rates as a risk factor for running related conditions and injuries.

4.5.1.4. Knee joint stiffness.

Knee joint stiffness is often defined as the change in sagittal plane joint angle in response to an applied joint moment from initial ground contact to peak knee joint flexion (Farley, & Morgenroth, 1999). Although mean stiffness has often been calculated throughout the entire stance phase, it is suggested that this method may overlook the possibility of differing stiffness values throughout the impact and active phase of gait (Butler et al., 2003). As such, for the purpose of this investigation, knee joint stiffness was calculated as the change in joint moment divided by the change in joint angle, from initial ground contact to peak knee flexion. In the neutral condition, results among the mature females showed knee joint stiffness to be significantly higher than that produced by the younger group. Despite this, no significant difference was displayed in the peak knee flexion angle between the two groups. However a significant difference was seen in the initial knee joint angle, with mature females showing higher angles of flexion at ground contact. It could therefore be considered that the increase in knee joint stiffness was in part, due to the increased knee angle at contact; reducing the overall knee excursion. This supports the previous research of Farely and colleagues, who suggested that joint stiffness may be primarily influenced by the initial knee joint angle (Farley et al., 1999), and Dixon and colleagues, who suggested that increased knee flexion at ground contact causes a reduction in knee excursion range and an increase in knee stiffness (Dixon et al., 2010).

Prior to normalising for body weight, the knee joint stiffness values ranged from 5.84 to 17.48 Nm/degree among the mature, and 4.04 to 12.09 Nm/degree among the younger group. These values fall in line with those presented in the literature, including an average of 4 Nm/degree for participants walking, and those ranging from 17 to 24 Nm/degree with increasing running speed (Dixon et al., 2010; Kuitunen et al., 2002). Within this investigation among female runners, the mature group was shown to display significantly greater values of knee joint stiffness than the younger group during running trials in a controlled neutral shoe. This suggests that an effect of age exists for this biomechanical variable; a notion also supported by Lark and colleagues (Lark, Buckley, Bennett, Jones & Sargent, 2003). In an investigation, Lark and co-workers showed a significant increase in joint stiffness among more mature men compared to younger men. Although these differences were predominantly seen at the ankle joint, during stepping motions, it is

suggested that muscle control mechanisms in response to activity change with age (Lark et al., 2003).

An increase in joint stiffness acts to maintain stability of the joint during gait, and reduce the movement excursions of the lower extremity. However, negative consequences have also been shown, with an increase in torsional stiffness associated with deleterious effects on the integrity of the cartilage. According to Griffin and Guilak (2005), increased antagonistic forces that produce stiffness often result in increased joint compressive forces, and increased risk for conditions such as osteoarthritis of the joint. Although not significant, the knee joint moments were higher among the mature group (young to mature 143.04 Nm to 154.56 Nm), where as the peak knee flexion angles were similar (young to mature 28.9^{o} to 32.3^{o}), suggesting that the observed differences in stiffness were contributed to primarily by the differences in initial knee angle and a slight increase in joint moment. This suggests that higher loads for the mature group acted over a similar range of articular surface, which could initiate cartilage degradation among the mature females (Zeni and Higginson, 2009).

As suggested previously, an increase in knee joint stiffness could, in part, be due to an increase in tightness of the hamstring muscles that is common with increasing age. Additionally higher values of knee joint stiffness could be contributed to by passive restraint caused by ligaments, joint geometry, friction between cartilage surfaces, and the load on the joint caused by compressive forces resulting from gravity and the muscles acting on the joint (Lephart, & Fu, 2000). Therefore, with a lack of knowledge regarding these features it is suggested that the values produced by the females enable only inferences on the exact stiffness of the knee joint of each runner in the current study.

The suggested association between knee joint stiffness and osteoarthritis has commonly been a retrospective one, with high magnitudes of joint stiffness often demonstrated among symptomatic diagnosed patients (Zeni et al., 2009; Dixon et al., 2010). However, with higher values demonstrated for a symptom free group of females, it is suggested that this biomechanical variable could be a risk factor for the development of debilitating conditions, such as osteoarthritis at the knee joint.

4.5.2 Hypothesis Two: A motion control shoe.

4.5.2.1. Effectiveness of a motion control shoe to alter kinematics during running.

The results from this investigation partially support the second hypotheses, as significant changes were observed in two variables with footwear interventions. When compared to the neutral shoe, the running trials performed in the motion control shoe illustrated significant reductions in the peak rearfoot eversion angle and the peak knee internal rotation angle among both groups of females. Furthermore, larger reductions in peak rearfoot eversion and knee internal rotation were found for the mature compared to the young group. Among the mature group, peak rearfoot eversion decreased by an average of 4.9^{o}, compared to a 2.5^{o} decrease among the younger group. Among the mature females, individual analyses illustrated nine from the ten participants to show a reduction in rearfoot eversion with the motion control shoe. Participant 5 however showed an angle of 5.7^{o} ($\pm$ 1.05) with the neutral and 7° ($\pm$ 2.66) with the motion control shoe. Similarly, among the younger group nine from the ten participants illustrated a decrease in rearfoot eversion with the motion control shoe with participant 10 showing an increase of 2.5^{o}. From the twenty two participants tested, 90.9% showed a decrease in rearfoot kinematics with the motion control shoe. In addition, the effectiveness of the motion control shoe in altering rearfoot kinematics was further supported by the lack of significant difference in the rearfoot eversion angle between the younger group and the mature females in the motion control shoe ($p>0.05$), suggesting a greater change in eversion angle among the mature group of runners.

Many running related injuries have previously been associated with malalignment of the skeleton during movement, due to the uneven distribution of forces across the joints (Nigg, 2001). As such, alignment of the musculoskeletal system has been one of the main functions of running shoes, inserts and orthotics (Nigg, 2001). The motion control shoe used within this investigation incorporated differing support on the medial and lateral sides, with an increase in arch height and sole hardness on the medial longitudinal arch. Biomechanically, the foot is represented in the current study as a rigid structure; therefore reduced medial movement of the mid foot is represented by the movement of the rearfoot.

As all mature females were categorised as heel-strikers, at ground contact the calcaneus strikes the ground on the lateral side and quickly progresses through to rearfoot eversion (Richards, 2008). As forward progression occurs, maximum subtalar pronation occurs to a position where by the body weight is positioned over the foot (Dungan, & Bhat, 2005). Although this motion is an essential movement associated with gait, the higher angle of rearfoot eversion shown among the mature group could increase risk of running-related overuse conditions in the lower limb (Yamashita, 2005).

The purpose of the "motion control" aspect in the running shoe is to reduce the frontal plane motion of the heel into eversion, and limit the ability of the longitudinal arch to collapse during the stance phase of gait (Rose et al., 2011), influencing the movement of the foot and orientation of the subtalar axis (Stacoff et al., 2000). Therefore, due to the coupling mechanism between the foot, ankle and tibia, the reduction in rearfoot eversion with the motion control shoe among the participants in this investigation was as expected, with all females showing a significant decrease in rearfoot eversion with the motion control shoe ($p<0.05$). Although not equivocal in the research literature, these results among the females do fall in line with much previous research. Cheung and Ng (2007) demonstrated a motion control shoe to reduce rearfoot eversion by 4° and Clarke (1983) similarly showed an increase in midsole hardness of a shoe to significantly reduce rearfoot motion and subsequent pronation. Perry and Lafortune (1995) demonstrated a motion control shoe insert with a 5° medial wedge, to significantly reduce rearfoot eversion by an average of 6.7^{o} and Eng and Pierrynowski (1994) illustrated a decrease in transverse and frontal plane movement at the ankle and knee to reduce with motion controlling foot orthotics during the early stance phase of gait. Both of these latter studies however investigated the effect of orthotics.

In contrast to the evidence provided in the previous paragraph, the positive effect of footwear has not been shown to be unanimous within the research literature (Butler et al., 2007). Results from a study involving bone pins placed in the calcaneus, tibia and femur during gait, showed little alterations in kinematics with different shoes and orthotics during running (Stacoff et al., 2000). This result queries the claim that footwear can align the skeleton during movement. This was later discussed by Nigg and colleagues, who suggested that the neuromuscular system is programmed to adopt a minimal resistance movement path, whereby appropriate muscles will act in response to an intervention that

changes this path (Nigg et al., 2001). However it could be speculated that the high eversion angles demonstrated among the mature females have developed as a result of age, and are therefore producing movements that are different from the original movement path when the females were younger. As such, the ability of footwear to change the lower limb kinematics among mature females as seen in this investigation could be due to the recent increase in rearfoot eversion among this group. This notion is further supported by the greater decrease in rearfoot eversion among the mature compared to the younger female runners. Further longitudinal research is required in this area to assess the change in kinematics over time, and the subsequent ability of footwear to alter the biomechanics during running.

Alongside reductions in rearfoot eversion, the motion control shoes also significantly reduced knee internal rotation among the mature females ($p<0.05$). This was however not similar for the younger females, with a non significant difference of only 1.7^{o} ($p>0.05$). Peak knee internal rotation angle represents the peak rotation of the tibia relative to the femur about a longitudinal axis embedded within the tibia. An internal rotation angle of 0° indicates the parallel position of the frontal plane of the tibia relative to the medial-lateral axis of the femur. As shown among the female runners, when running in a supportive shoe, the ability of the rearfoot to move into eversion is limited. As such, the normal movement of the tibia and knee is consequently reduced. This is due to the axial coupling between the foot segment and the tibia bone, via the anterior talofibular and superior deltoid ligaments (Huson et al., 1986). Additionally, when the subtalar joint everts, the large range of motion is permitted at the joint and axis, permitting the shock absorbing mechanism of this movement. However, when in inversion, or before eversion is achieved, the midtarsal and subtalar joint are locked, reducing the movement transmitted to the tibia and proximal joints of the lower limb (Huson, 2000).

This effect of motion control shoes on knee movement was supported by Butler and colleagues (2007), who demonstrated a motion control shoe to significantly reduce tibial and knee internal rotation. However, surprisingly there was no change in the subtalar joint motion with the motion control shoes, which contradicts the previous theory that rearfoot motion and tibial rotation are linked. It is speculated that as the support was centred in the midsole of the motion control shoe, the effect on the rearfoot may not have been as influential (Rose et al., 2011). These results were similar to those produced by Nigg and

colleagues, who found that medially posted moulded orthotics, had a negligible effect on rearfoot motion during running, although a significant reduction was shown in the rotation of the tibia, with a 31% reduction shown in the mean angle (Nigg et al., 1999). In a more recent research study by Rose and colleagues a motion control shoe was once again shown to significantly reduce tibial rotation among a group of healthy runners (mean difference of 1.4^{o}). In this research, Rose and colleagues did not however look at the effect on rearfoot motion (Rose et al., 2011). If tibial and knee internal rotation are taken as a coupled mechanism, (Kirtley, 2006) these results seem to strongly support those from this investigation among mature female runners.

Although the mature females showed a significant decrease in knee internal rotation angle with the motion control shoe, there was still a significant difference between the peak angle produced by the two groups. This suggests that although the motion control shoes were effective, the reduction in angle was not sufficient to cause the mature females to run with rotation values similar to a group of younger runners. As such, results imply that although the current level of support in the tested motion control shoe may reduce knee internal rotation among the mature females, the level of support may not be sufficient to alter the increase in rotation that appears to occur with age among female runners.

Although the inclusion of a motion control shoe did not significantly alter ankle dorsiflexion, knee abduction or loading rate among mature and young female runners, results from this study suggest that wearing a motion control shoe such as the Supernova Sequence could significantly reduce the likelihood of mature females developing an overuse injury or condition related to eversion and subsequent knee internal rotation. It has been shown in this investigation that mature female runners as a specific group react as intended to the motion control element for the running shoe model tested.

4.5.2.2. Additional factors influencing the effectiveness of a motion control shoe.

It has previously been suggested that patient or shoe characteristics may mediate the effect of motion control shoes or footwear interventions on the biomechanics of running gait, suggesting that certain individuals may be more likely to respond (Hinmann & Bennell, 2009). Therefore an additional consideration when investigating the efficacy of a motion control shoe in modifying biomechanical parameters of gait is the level of support in the shoes worn by each participant on a regular basis. This is supported by Hintermann and Nigg (1998), who suggested that an increase in habituation time to additional support in trainers causes an incremental increase in the effectiveness of the orthotics to alter gait. Therefore, although not assessed, if motion control shoes were worn by mature females on a regular basis, the rearfoot eversion angle may be more adaptable with the inclusion of this footwear type in the study, when compared to those who wore neutral shoes regularly. Additionally, the manufacturer brand and design of the regularly worn trainers may alter each individual's adaptation to the Adidas shoe provided within this study. This difference may be a cause for the inconsistent reduction in knee internal rotation among the younger females, or the lack of change in knee abduction angle or loading rate of impact force with the motion control shoe among either group.

Similarly, the gait inherent to each individual female runner may mediate the biomechanical responsiveness to a motion control shoe; however this has not been extensively reported in the literature. One study by Lidtke and colleagues (2006) did however show an increase in baseline rearfoot eversion to be associated with an increase in reduction of the knee external adductor moment with a lateral wedge footwear intervention. As such, it could be considered that the higher levels of rearfoot eversion shown by the mature females could in part, provide explanation for the greater reductions in rearfoot eversion and knee internal rotation angle seen with the motion control shoe among this group.

4.5.3. Hypothesis Three: The knee external adductor moment.

The results from this investigation did not support hypothesis three. The motion control shoes did not significantly increase knee abduction angle, and did not alter the external knee adductor moment in either group. It was expected that the medial support in the motion control shoes would significantly increase the varus (abducted) position of the knees, due to a lateral tilt provided by the mid sole of the shoe. This would reduce the possibility of the knee moving into valgus (adduction), and cause an increase in the lateral movement of the knee joint during stance. Similarly, the medial support provided by the motion control shoe would potentially move the line of action of the ground reaction force and centre of pressure medially, and increase the varus position of the legs, thereby increasing the moment arm at the knee (Hunt et al., 2006).

The knee external adductor moment has been linked to symptoms, progression and initial development of medial knee joint osteoarthritis. Due to this strong association, this biomechanical variable has recently become recognised as a non-invasive method for investigating knee joint forces, and acts as a valid and reliable representative of medial to lateral load distribution at the knee joint (Radzimski, Mundermann, & Sole, 2011). As such, techniques to reduce the knee external adductor moment during gait have received a vast amount of interest, and footwear modifications have been viewed as a potential conservative management of knee joint osteoarthritis (Radzimski et al., 2011).

The non significant change among the female runners lies in contrast to a sample of results presented in the literature, that investigated motion control shoes and medial knee joint loading. One study investigated the difference between barefoot running, and running in a motion control shoe (Brooks Adrenaline) and showed an average increase of 38.33 % in the knee external adductor moment among healthy individuals (Kerrigan et al., 2009). Similarly, another study looked at the effect of a current stability (motion control) shoe and a mobility shoe (indicative of barefoot) on the knee external adductor moment (Shakoor et al., 2008). Results also indicated that when compared to the barefoot condition and the barefoot simulator shoe, the motion control shoe instigated a slight increase (8%) in the external adductor moment. However these results were demonstrated among a group of symptomatic OA patients during walking trials. Schmalz and colleagues also investigated the effect of a medial wedge on the walking gait of healthy

individuals and showed a 9.26 % increase in knee external adductor moment when compared to a non wedged insole (Schmalz et al., 2006).

As such, it appears that the results produced by the mature females fall in contrast to those presented in the literature. It is speculated that as the mature females had a high level of rearfoot eversion in the neutral condition, the motion control aspect of the shoe provided only enough support to bring the foot back to an ideal alignment of the subtalar joint. Similarly, as the peak external adductor moment occurred during late stance, it is possible that the medial support based in the heel and mid foot was not adequate to alter the peak moment produced.

According to Hunt and colleagues, the knee external adductor moment is more dependent on the length of the moment arm than the magnitude of the ground reaction force, suggesting that the levels of support in the shoe designed to alter alignment of the lower limb will have greatest effect on the moment (Hunt et al., 2006). This is supported by an additional study, where a 1% increase in moment arm length was shown to cause a 6.5 times increase in radiographic progression of medial knee osteoarthritis (Hurwitz et al., 1998). This is indirectly supported by the results produced by the female runners in the current study, as no significant change was demonstrated in the moment arm between the two footwear conditions, consistent with the lack of difference in the knee external adductor moment ($p>0.05$).

According to Kakihana and colleagues, changing the component of footwear will alter the position of the centre of pressure, and subsequent moment arm at the knee; in one study, a lateral wedge in the shoe caused a lateral shift in the centre of pressure, decreasing the knee external adductor moment (Kakihana et al., 2005). Additional studies have investigated varying stiffness levels of the shoe, and consistently showed stiffness of the lateral as opposed to the medial sole to instigate a reduction in knee external adductor moment. Fisher demonstrated a 7.2% decrease in knee external adductor moment with an increase in lateral sole stiffness among a healthy population, and Erhart found an increase in lateral support to decrease the moment by an average of 6.2% among symptomatic OA patients (Fisher et al., 2007; Erhart et al., 2008). As such, it is suggested that a footwear adaptation including lateral support in the shoe will be expected to have the greatest effect on the knee external adductor moment during stance.

Since the external adductor moment is commonly associated with osteoarthritis development at the medial knee, the high knee external adductor moment for the mature compared with the young runners is potentially detrimental. Therefore, to reduce this moment among mature females, a footwear intervention is advised, with increased support on the lateral sole.

4.5.4. Hypothesis four: Muscle strength.

Hypothesis four suggested that the mature females would display reduced quadriceps strength when directly compared to a group of younger runners. Results from the current investigation however did not support this expectation. As illustrated in Table 4.8, although an average of 10.66 Nm difference was demonstrated between the two groups, this difference was not significant ($p>0.05$). This suggests that the concentric strength of the quadriceps muscles was not affected by age, between a mean age of 21.2 and 49.7 years.

Although it was hypothesised that muscle strength would be lower for the mature group, it is possible that the average age of the mature females was not great enough to warrant a difference in strength. According to Larsen and colleagues, muscle strength is maintained between the ages of 30 to 50 years, however from 50 to 70 years there is a 30 % decline in force generating capacity within the muscles (Larsson et al., 1979). As such, as the average age of the mature group was below 50 years, the decline in strength may not have been initiated.

As the age of the mature group spanned 20 years, a regression analysis was performed to assess whether an effect was seen for age within this particular group. Although the scatter diagram (Figure 4.21a) appeared to graphically illustrate a linear relationship, the results were not significant ($p>0.05$). As shown in the scatter diagram, one participant seemed to produce results that did not follow the linear trend line. Participant 6 within the mature group produced a muscle strength maximum of only 49.2 Nm, a value lower than what would be expected for this female's age if following the linear relationship. It is likely that this one subject's data may have resulted in a non significant regression result.

Although not assessed in this study, a reduction in muscle strength is often associated with a decline in muscle mass and muscle fibre size. According to Young and colleagues, among mature females, the cross sectional area of the quadriceps muscle explained 44 % of the difference in strength compared to a younger group (Young, Stokes, & Crowe, 1984). This was later supported by Lambert and colleagues who, in a review of the literature, suggested that cross sectional area of muscles is the factor that most influences muscle force production (Lambert, & Evans, 2002). Therefore, based on the results produced by the mature group, it is suggested that the cross sectional area of the quadriceps femoris muscle was not significantly different between the two groups.

Figure 4.21b illustrates the relationship between muscle strength and body mass. Although a positive relationship was shown in the graph, results from a regression analysis proved the relationship to be non significant. This is contrary to previous research, whereby increased weight among females has commonly been associated with decreased muscle strength (Slemenda et al., 1998). However it is noted that increased body mass in this researcher's investigation was related to obesity. Among the mature females, the maximum body weight was 60 kg; a value that would not categorise any mature female as obese. This was as expected, as all females were categorised as runners; each training three times in a week.

The concept of regular exercise is an additional factor that may explain the lack of significant difference in the muscle strength between the two groups of females. As a sedentary lifestyle is a common catalyst for the relationship between age and strength decline, it is suggested that as all the mature females within the current study train (running) for the same weekly duration as the younger group, the muscle mass among the former group may be maintained for longer. This is supported by Eyigor (2004) who showed a progressive resistive exercise programme to be effective in increasing flexor and extensor peak torques among a group of participants aged 40 to 70 years. It is noted here that all participants had some stage of osteoarthritis, further supporting the theory that strength may reduce the onset of osteoarthritis among the mature members of the population (Eyigor, 2004).

This association between osteoarthritis and muscle strength has been reported in the literature. It has been proposed that reduced quadriceps strength is associated with increased osteoarthritis risk. In a prospective study by Slemenda and colleagues, an 18%

reduction in quadriceps strength at baseline was associated with the eventual development of osteoarthritis among females (Slemenda, Heilman, Brandt, Katz, Mazzuca, Braunstein, & Byrd, 2004). However, no mention of the activity levels of these females was presented. It is suggested that although the mature female runners may be at an increased risk of osteoarthritis development, a reduction in muscle strength does not appear to be a factor.

4.6. Conclusion and Future Direction.

In light of the results presented, it is apparent that the application of a motion control shoe significantly alters certain gait parameters inherent to the mature female runner (40 to 60 years). Trials performed in the controlled neutral running shoe confirmed findings from the previous study, and showed mature females to exhibit a significantly different running gait pattern compared to younger females. Among the mature group, the motion control shoe was shown to significantly reduce the magnitude of peak rearfoot eversion and knee internal rotation; however the peak knee external adductor moment was not changed. In light of the results of this investigation in conjunction with previous relevant literature, it is suggested that although the current motion control shoe reduces certain biomechanical differences between young and mature runners, further footwear adaptations should be made to adequately reduce other variables associated with the gait of mature females.

4.6.1. Future Research and Intervention shoe.

Excessive joint movement in specific directions during gait has been shown to be a risk factor for the development of conditions and injuries among runners. An intervention shoe designed to control the gait of mature female runners is suggested.

Increased arch support has been shown to decrease rearfoot eversion and knee internal rotation. Therefore, this component of medial support is essential. The current motion control shoe (Supernova Sequence) was designed with a simplified arch support system. However, a flattened medial arch is likely to cause negative effects on the knee joint. A low support medial arch during running increases the risk of soft tissue injuries on the medial side of the lower limb (Norris, 2004). Furthermore, decreased arch support on the medial side causes the foot to roll medially during running, causing an uneven distribution of load, and instigating excessive internal tibial rotation and subsequent internal rotation of the knee joint (Mirkin, 2005). In addition, it has been suggested that increasing age can cause slight reductions in arch support, and increased flattening of the medial arch, due to increased laxity of the plantar ligaments supporting the arch of the

foot which occurs with age (Moore, 1992). Finally, in an article presented in Runner's World Magazine, a review about the Supernova Sequence Women, indicated a common theme that females felt the medial arch was not supportive enough (Runner's World, Jan 2010).

In addition to arch support, the inclusion of an increased stiffness on the lateral sole is suggested for the intervention shoe designed for mature females. As one of the variables most commonly associated with osteoarthritis development at the medial knee, the high knee external adductor moment for the mature compared with the younger runners is important. A high peak knee external adductor moment has been associated with the severity, rate of progression, and treatment outcome of medial compartment knee joint osteoarthritis (Baliunas et al., 2002; Miyazaki et al., 2000). Subsequently, researchers have suggested a lateral wedge as an intervention to decrease this moment. Although such methods have been shown to reduce the adductor moment, the use of a lateral wedge has a major limitation in its cause of discomfort (Kerrigan et al., 2002). Therefore, a variable stiffness shoe is advised, with arch support combined with increased stiffness on the lateral sole. This has been previously shown to decrease the knee external adductor moment during walking to the same extent as the wedge in both healthy and osteoarthritic subjects, with no noted reports of discomfort (Fisher et al., 2007; Erhart et al., 2008).

Although the running gait changes that occur with age have not been extensively documented with respect to females, certain differences are still applicable to the mature female runner. According to Whittle (2003) advancing age causes changes to occur to the musculoskeletal system that can directly and indirectly affect gait. A main structural change is the reduction in the thickness of the plantar fat pad of the foot (Frey, 1997). Repeated impact as occurs during running could therefore increase the risk of injury or pain in this area. Increased cushioning in the shoe sole may counteract this possible effect.

Although abundant on the shelves of running shoe shops, current motion control shoes do not control for all components of gait that are related to injury occurrence. Mature female runners have demonstrated certain gait parameters that could predispose them to injuries and conditions at the knee (knee external adductor moment), and are unaffected by motion control components in shoes. Therefore, in light of the results of this investigation in conjunction with previous relevant literature, it is suggested that certain adaptations are

made to the current motion control shoe. This will provide a shoe design specific for the mature female runner, aimed at reducing the currently high incidence of running related conditions.

Chapter 5.

"The effect of a lateral wedge and arch support orthotic on the running biomechanics of mature females."

5.1. Introduction and Review of Literature.

5.1.1. Review of previous study findings and conclusions.

Mature female runners have been shown to run with increased rearfoot eversion and knee internal rotation, as well as significantly greater knee external adductor moments compared with younger females (Lilley et al., 2011). In high magnitudes, all three components of gait have previously been associated with the predisposition to, and development of, running related conditions at the knee joint. Furthermore, the use of a motion control shoe has been demonstrated as successful in reducing the rearfoot eversion and knee internal rotation movements (Lilley et al., 2011). However, the knee external adductor moment, frequently associated with knee injury and debilitating conditions (Miyazaki et al., 2002), was not shown to be influenced by this shoe design. This chapter considers alternative approaches to reducing knee external adductor moment among mature female runners.

5.1.2. Rearfoot eversion, knee internal rotation and forefoot abduction.

Rearfoot eversion is an essential component of gait, dominant in early stance and required for shock absorbency. However excessive rearfoot eversion has often been related to injury occurrence. High magnitudes of eversion of the rearfoot suggests a hyper-flexible and unstable foot (Cheung et al., 2007), and previous studies have indicated a link with increased risk of Achilles tendon injury, plantar fasciitis, and shin pain (Clement et al.,

1984; Stacoff et al., 2000). Abnormal eversion of the rearfoot occurs when the degree of eversion is excessive (above 15°), or when pronation occurs when the foot should be supinating (Kingston, 2000; Brukner and Khan, 2006). Furthermore, an excessively everted rearfoot tends to internally rotate the lower limb during weight-bearing, through the transfer of movement proximally along the limb. This increases the demands and loads placed on the anatomical structures of the leg (Brukner and Khan, 2006) and as such, the link between excessive eversion and knee rotation has been shown to increase the risk of knee conditions and pain (Stacoff et al., 2000). Internal rotation of the tibia and knee joint causes lateral subluxation of the patella, and a change in alignment of the patella tendon. These contribute to increased tightness of the iliotibial band, and overall patellofemoral dysfunction during the early stance phase of gait (Brukner and Kahn, 2006).

As previous research has shown rearfoot eversion and knee internal rotation to be excessively high among a group of mature females, it has been suggested that the running gait of this specific group may predispose to injuries and conditions at the knee joint. The previous study demonstrated that the use of an existing motion control shoe with increased arch support however significantly reduced both variables among mature female runners. Therefore it has been suggested that this intervention may be suitable for this specific group.

Although not assessed in the previous studies, forefoot abduction has previously been associated with excessive values of subtalar joint pronation, and therefore may be high among mature female runners. As described, the movement of the midtarsal joint is restricted due to the ligaments crossing the joint; the predominant movement permitted is that of abduction and adduction of the foot (Perryn, 1992). Adduction involves medial flexion of the anterior part of the foot, and is one of the movements associated with inversion, whereas abduction involves lateral flexion associated with eversion (Jenkins, 1998). The initial rearfoot inversion at ground strike is accompanied by adduction of the midtarsal joint. During the loading response of pronation, abduction of the midtarsal joint occurs (Whittle, 2003). However, although it is clear that a small degree of both abduction and adduction are required throughout the gait cycle, deviations from the normal magnitude of these angles may lead to degeneration of the joints and subsequent development of debilitating conditions. Forefoot abduction describes the supinated

position of the metatarsal heads in relation to the hindfoot when the subtalar joint is fixed in neutral (Root, Orien., & Weed, 1977). During gait, excessive forefoot abduction tends to induce increased pronation, to enable the medial metatarsal heads to contact the ground (Alonzo-Vaquez, Villarroya, Franco, Asin & Calvo., 2009). As such, this forefoot movement has been related to the injuries associated with excessive subtalar joint pronation (Alonzo et al., 2009). Although not assessed with a motion control shoe in the previous study, it may be interesting to assess the effect of footwear interventions on this parameter of gait. Due to the association with rearfoot eversion it is suggested that a medial wedge may be suitable to reduce this movement among mature female runners.

According to Kernozek and Richard (1990), excessive abduction during the stance phase of gait can lead to a lowered longitudinal arch of the foot (flat foot), and increased pronation. This in turn is then related to increased rear foot motion, excessive rearfoot eversion and greater risk of load related conditions developing at the knee joint. This notion was later supported by Myerson, (1999) who stated that flat foot deformity during the stance phase of gait was associated with increased abduction of the foot. Furthermore, it was suggested that abduction of the midtarsal joint was related to increased lateral displacement of the foot, lower limb malalignment, and subsequent medial displacement of the resultant force acting through the knee joint.

5.1.3. The knee external adductor moment, medial knee joint loading and osteoarthritis.

The knee external adductor moment, acting in the frontal plane during stance phase, has been shown to be consistently high among mature female runners (Lilley et al., 2011). However, current motion control shoes with medial support have been shown to have little effect on this potentially harmful component of running gait (Lilley et al., 2011).

As described in Chapter 3, lower limb alignment of the hip, knee and ankle, contributes to the distribution of load by proportionately dividing it across a load bearing axis, between the medial and lateral components of the joint (Hunter et al., 2005). This axis is represented by a line drawn from the mid femoral head to the centre of the ankle, and during weight-bearing in neutral alignment, the medial compartment is known to

accommodate between 60 and 70% of the entire load experienced at the knee joint (Jenkyn et al., 2008).

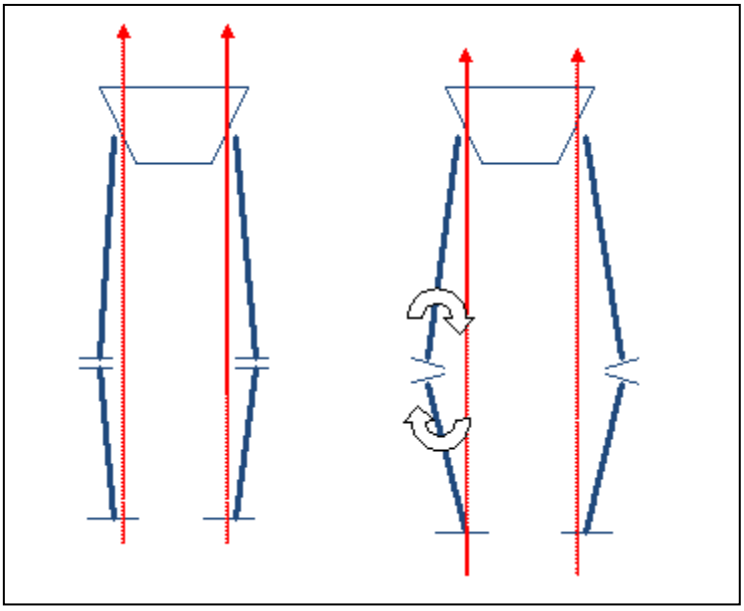


Figure 5.1. The external adductor moment acting at the knee joint (right) compared with "normal" alignment (left). View from the front.

It is widely accepted that both the magnitude and distribution of biomechanical forces are associated with the development of degenerative conditions at the knee joint (Andriacchi et al., 2004). When a knee external adductor moment acts at the knee joint, the relative load on the medial compartment is increased, and the risk of the initiation and progression of these conditions increases (Miyazaki et al., 2000; Zhao et al., 2006). High magnitudes of this variable experienced during walking and running gait have therefore been shown to increase the development of degenerative conditions such as osteoarthritis of the medial compartment (Sharma et al., 1998; Hurwitz et al., 1998; Jenkyn et al., 2008).

As described in Chapter 2, osteoarthritis is a degenerative joint condition, and the most common form of arthritis to affect people of all ages. It is however most common among the more mature members of the developed world (Hinton, Moody, Davis & Thomas,

2002). This condition occurs as increased loads on the medial compartment lead to degeneration of the articular cartilage surrounding the femur and tibia (Foroughi, Smith & Vanwanseele, 2009). Early stages are defined by evidence of thinning cartilage, bruising of the underlying bone (oedema), and the development of osteophytes and sub articular cysts (Gormley & Hussey, 2005). However, as articular cartilage is not innervated, these radiographic signs are often not married with clinical signs and symptoms of parity (Gormley & Hussey, 2005).

5.1.4. The effect of footwear on the knee external adductor moment.

Orthoses are a means by which to alter the alignment of the foot and lower limb during static stance and gait. As such, a medially or laterally inclined wedge is bracketed under the same heading as an orthotic, with the intention of modifying the alignment of the lower limb during locomotion.

Over the past two decades, research has been conducted to investigate the effect of modified footwear as a potential conservative management strategy for knee osteoarthritis. Although the majority of this research has focussed on diagnosed symptomatic patients during walking, interventions that are proven to reduce this moment, are a key step to potentially managing symptoms and reducing rates of knee osteoarthritis.

Among healthy subjects, studies investigating the effect of footwear on the knee external adductor moment have predominantly involved subjects performing walking trials. However two studies have looked at symptom free subjects running at self selected speeds. Kerrigan et al (2009) investigated the effect of a motion control shoe on the knee external adductor moment, and demonstrated a 38.33% increase in moment among a group of males and females with a mean age of 34 years. In an additional study, Eslami et al (2009) recruited 11 males with a mean age of 27.9 years, and assessed knee external adductor moment among subjects running in sandals and sandals with a semi rigid orthotic. Results demonstrated a slight decrease in the knee external adductor moment among trials performed in the orthotic compared to the sandal. However these results

were insignificant with large standard deviations, and were attributed to differences in individual adaptation to footwear interventions (Eslami et al., 2009).

Although the above investigations have looked at medial support modifications, lateral wedging of shoes has been investigated more frequently with the aim to reduce the knee external adductor moment (Radzimski et al., 2011). One early study looked at the effect of a nickleplast lateral wedge extending the entire inferior border from hindfoot to forefoot, on the external knee adductor moment of 17 female participants (mean age 27.7 years). Participants wore their own shoes as the control, and then the orthotics were added providing the test trials. After walking at a self selected speed, a significant mean reduction of 6.76% was demonstrated in the peak knee external adductor moments in the lateral wedge condition (Crenshaw et al., 2000). This was supported in later investigations, where a 3 degree and 6 degree lateral wedge were shown to reduce the knee external adductor moment by 8.9% and 24.4% respectively among 10 participants (5 males, 5 females, mean age 25 years) (Kakihana et al., 2004), and a 14 degree lateral wedge reduced the moment by 5.6% among 6 males and 4 females aged 34 years (Shmalz et al., 2006), although this latter result was reported insignificant in the article. Kakihana and colleagues later performed two studies isolating males and females, and showed a 6 degree lateral wedge tested under the same conditions to demonstrate reductions of 9.38% and 11.33% among males and females respectively (Kakihana et al., 2005; Kakihana et al., 2007).

A later study also looked at the effect of increasing the speed of walking on changes in knee external adductor moment. Erhart et al (2008) demonstrated both a 4 degree and 8 degree lateral wedge to significantly reduce the knee external adductor moment among 9 females and 6 males (mean age 28.6 years), with an additional effect for speed. When three walking speeds were assessed, the addition of the 8 degree lateral wedge illustrated the greatest decrease in knee external adductor moments at the highest speed (slow, 17.3% reduction; fast, 19.1% reduction).

Therefore, although this did not look at the effect of running, results suggest that the reductions in the knee external adductor moment through shoe modifications may be greater when subjects perform running trials. Additionally, reductions in knee moment may demonstrate a gender effect, with reductions seemingly greater for female participants when compared to males.

With regard to the results achieved in the previous chapter, it is suggested that mature females may benefit from a range of control interventions in their running shoes. Although little research has investigated a multipurpose modification to running shoes, to my knowledge one study has investigated the effect of combining both lateral wedge and medial arch support on the gait of healthy volunteers. Nakajima and colleagues (2009) investigated the effect of a lateral wedge combined with arch support, on the gait of 11 males and 9 females (mean age 28 years) walking at a self selected speed. Results showed that the lateral wedge alone significantly decreased the knee external adductor moment by 8.8%, and the inclusion of the arch support decreased the knee moment by 7.5% compared to the neutral condition. Therefore, although the arch support produced a slight reduction in the percentage decrease, the knee external adductor moment was still significantly lower than in the control condition (Nakajima et al., 2009).

Similarly, Schmalz et al (2006) investigated the effect of footwear modifications on the walking gait of 6 males and 4 females, aged 34 years (±9). As well as investigating the effect of a lateral wedge, this discussed research included the addition of a semi-rigid ankle support to restrict movement in the frontal plane. Results indicated that when wearing a lateral wedge combined with the ankle support, the knee external adductor moment was reduced by 9.26%. However, although indicating a significant effect, the frontal plane support used by Schmalz and colleagues was obtrusive, and not applicable for the recreational use of healthy individuals.

The salfordinsole orthotic, designed by Dr Richard Jones, comprises lateral wedge technology with arch support incorporated. Research incorporates in the design process showed this orthotic to significantly reduce the knee external adductor moment among patients with mild and severe medial compartment knee osteoarthritis. Furthermore, it was also shown that the inclusion of the medial support significantly preserved the height of the medial longitudinal arch, thereby eliminating the consequential increase in subtalar joint pronation possible with an increase in lateral support. As such, this readily available orthotic is deemed appropriate for the mature female runners to reduce the currently high levels of rearfoot eversion, knee internal rotation angle and knee external adductor moment during running.

5.1.5. Aims and hypotheses.

In light of the combination of discussed literature and the results from the previous study, it is suggested that a range of orthotics are investigated, and the effect on the running biomechanics of mature females is analysed. The aim of this investigation was to assess the effect of both a medial and lateral wedge, and a combined orthotic with both medial and lateral support, on the running gait of mature females. Three hypotheses were developed.

1. During running, the medial wedge will produce a significant decrease in the rearfoot eversion and knee internal rotation angle among mature female runners, with no adverse effect on the knee external adductor moment.
2. During running, the lateral wedge will produce a significant decrease in the peak knee abduction angle and peak knee external adductor moment among mature female runners, with an accompanied increase in rearfoot eversion and knee internal rotation angles.
3. The orthotic device will reduce rearfoot eversion and knee internal rotation angles, and produce significant reductions in the knee external adductor moment among mature female runners.

5.2. Methods.

5.2.1. Pilot study.

A pilot study was conducted in order to finalise the test conditions appropriate for this study. Six wedges were tested, alongside one orthotic intervention. The wedges tested comprised a selection of 2 degree, 4 degree and 6 degree lateral and medial wedges spanning the length of the shoe, made from flexible plastic. These wedges were added to the shoe under the manufacturer footbed, to control for excessive movement of the rear and forefoot in the frontal plane of motion. The orthotic intervention comprised both medial rearfoot and arch support and lateral rear and mid foot support. This was tested to ensure no discomfort was experienced when running.

The wedges and orthotic intervention device were tested on a mature female (age 50 years), who volunteered for participation. This participant performed 10 running trials in each randomly selected condition, and provided verbal feedback on each condition. Three dimensional biomechanical data was captured using the Peak Vicon Motion Capture System, and the main variables of rearfoot eversion and knee internal rotation were assessed. Knee external adductor moment was calculated using inverse dynamics through MATLAB.

Results indicated no significant effect for both the 2 degree and 4 degree medial and laterally wedges ($p>0.05$). However a significant effect was seen in all three variables in both the 6 degree lateral and medial wedges, and the orthotic intervention ($p<0.05$). Additionally, no discomfort was experienced in either the 6 degree medial or lateral wedge, or with the orthotic intervention. As such, the test conditions deemed appropriate were the 6 degree medial and lateral wedges and the orthotic intervention.

5.2.2. Participant selection and KOOS.

Based on the decrease of 6.09 degrees in rearfoot eversion with a motion control shoe seen in Study 2, a sample size of 20 mature females would produce a power of 87%. Twenty able-bodied mature females were therefore recruited on a volunteer basis from a local women's running club (Women's Running Network). Of this group, ten had been included in the previous study. Females were recruited from the age bracket of 40-60 years, and selected volunteers ranged in age from 42 years to 59 years (Mean, 50.7, sd, 4.5). All volunteers described running as recreational, each with a minimum of 5 years running experience. Each female participated in at least three one hour sessions of running per week, which was standardised as all females ran on training evenings. Females had no history of orthopaedic or neurological ailments that could affect running gait, and were free from symptoms of a musculoskeletal injury at the time of testing. Each female's average 10 km run time was used as an indicator of running standard. All participants were within a 5% time range (60.9 ± 2.96 minutes).

As a prerequisite, prior to the start of gait analysis, each female completed a Physical Activity Readiness Questionnaire; PAR-Q form to assess participant statistics (age, height, weight) and to ensure all females were free from any major visual or balance pathology, or debilitating condition that could hinder performance (PAR-Q, Canadian Society for Exercise Physiology, 2002). Participants also completed a KOOS (knee osteoarthritis observation survey) to assess the presence or potential of osteoarthritis and injuries present at the knee joint (Roos et al., 2003). KOOS results were correlated with collected movement data to investigate potential links between knee osteoarthritis and lower limb biomechanics during running. A summary of demographic data and KOOS scores for both groups is illustrated in Table 5.1. The experimental procedures were approved by the University of Exeter research ethics committee.

5.2.3. Test conditions.

Within this study, the effect of four different shoe conditions was assessed. The shoes incorporated were Somnio Women's Pacemaker 2.0 (Somnio Inc., 2010), a neutral women's running trainer. These specific shoes were selected for this study due to the customisable feature; F.E.A.T (Functionally Engineered Adaptable Tricomponent System). This provided interchangeable foot-beds for the shoe based on individual arch height, and adjustable medial/lateral wedge inserts to accommodate for differences in biomechanics. Each part is presented in Figure 5.2.

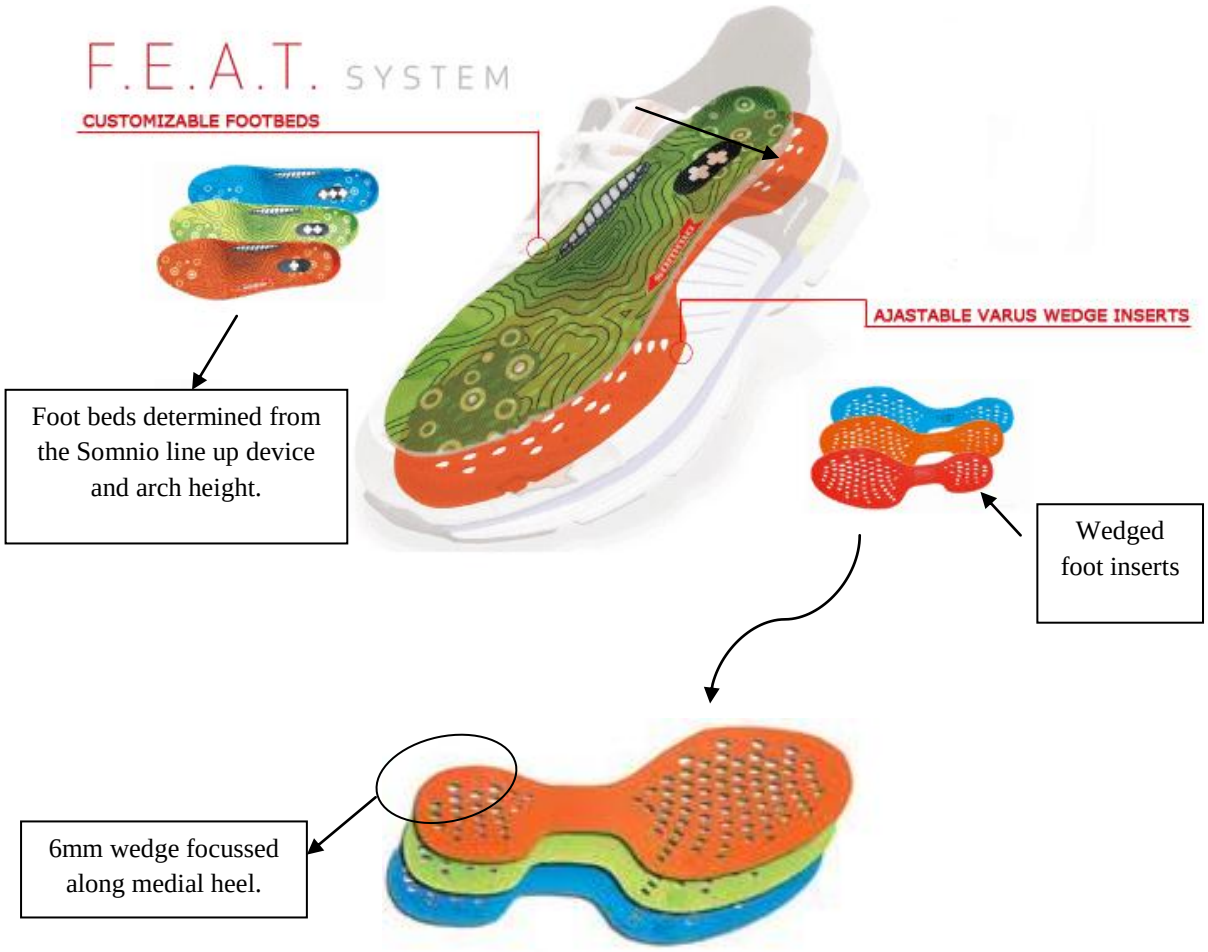


Figure 5.2. Illustration of the trainers, foot beds and wedged inserts added to the shoes. Source: Somnio FEAT system, Inc, 2010.

Within this study, the foot-bed added to each female's shoe was determined using the Somnio Line-Up Device (Somnio, Inc. 2010) (Figure 5.4). The procedure for the line-up device arch height assessment is detailed in the diagram below (Figure 5.3). This assessment enabled the specific foot bed to be selected for each participant, adjusting for "normal, low and high" arch types (Somnio Inc, 2010).

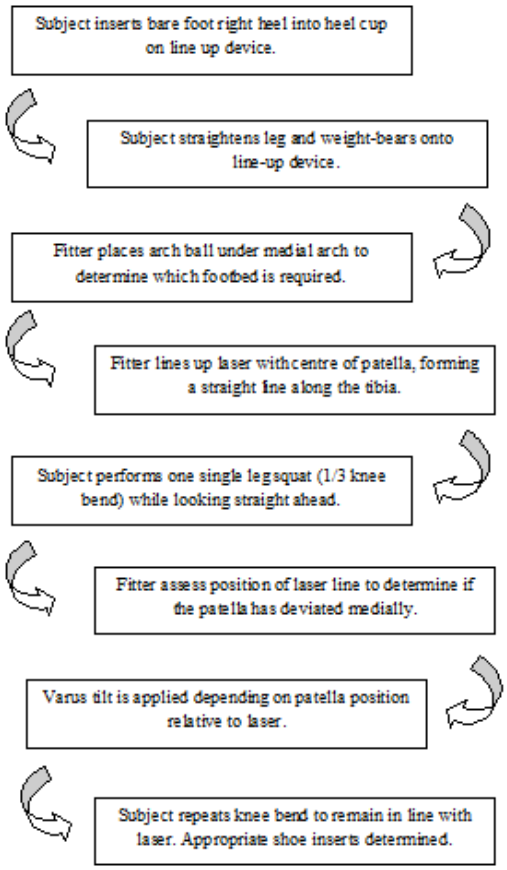


Figure 5.3. Somnio Line-Up Device Procedure, as adhered to in this study. (Information from Somnio Inc, 2010).

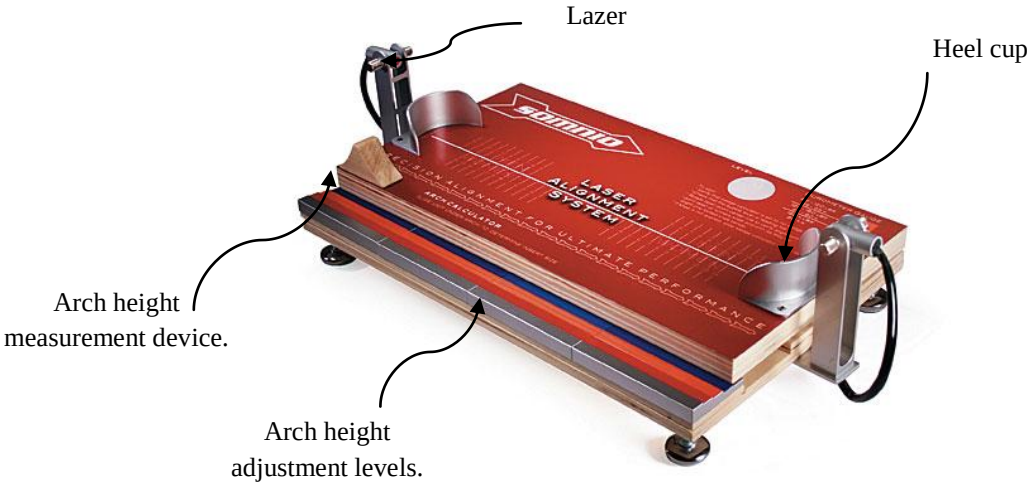


Figure 5.4. Annotated Image of the Somnio Line Up Measuring device and different footbeds.

The results from the arch height assessment and laser line-up test are presented in section *5.3.1.2*. Arch height determined which foot-bed was required. Each individual's selected foot bed was then inserted into the shoe, and this was labelled the neutral (control) condition for each female. As each individual was assessed with the same medial and lateral inserts, the results from the line-up test were noted, and later correlated with the kinematic data collected for each individual.

The medial and lateral inserts were wedged with a 6 mm incline on the either the medial or lateral sides respectively, focussed in the heel. These were placed in the running shoes under the footbed, and were tested as separate conditions. The final condition was an orthotic device based on the results from the previous study, and included a lateral wedge as well as rearfoot control. These orthotic devices were made from a thermoplastic elastomer, with silver-based anti-microbial agents embedded in the sole (Salfordinsoles, Incorporating Lateral Wedge Technology, Inc, 2008) (Figure 5.5). The orthotic devices were added to the shoe in place of the foot bed as an additional test condition.

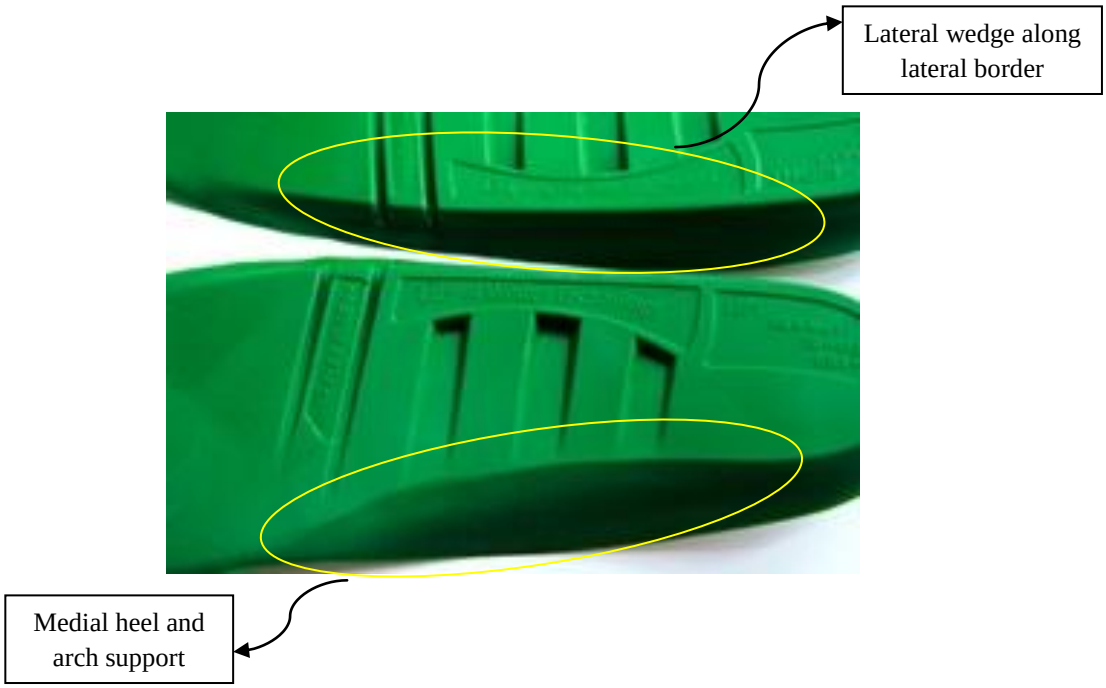


Figure 5.5. Image of the orthotic intervention, incorporating arch support and lateral wedge technology. Source: Salfordinsole™

5.2.4. Experimental procedure for assessment of running gait.

Three dimensional analysis of human gait occurred using an 8 camera motion capture system (Vicon Peak, 120 Hz, automatic, optoelectronic system; Peak Performance Technologies, Inc., Englewood, CO.) synchronised with a single floor mounted force platform (960 Hz, AMTI, Advanced Mechanical Technology, Inc., Massachusetts). Data were synchronised within the Vicon software using initial foot strike as an automatic event detection (vertical force > 10N). The 8 cameras were positioned in an oval shape surrounding the force platform, located in the middle of a 10 meter runway. The capture volume (1.0 m in length, 0.5 m in width, and 0.75 m in height) encompassed the right lower limb motions of each female during the running trials.

5.2.4.1. Marker placement and static trial.

The model of the right lower limb, on which the methodology is based, is that of a series of rigid linked bodies. These move in space with six degrees of freedom, describing the three lineal translations, and three angular rotational movements of each segment (Giannini et al., 1994). In order to obtain the description of this movement, each segment is associated with a rigid configuration of spherical markers, fixed in known locations. In this study, eleven reflective markers were attached to the right lower limb of each female, using a modified version of the model presented by Soutas-Little (Soutas-Little et al., 1983). This was the same model used in the previous chapters; therefore location of marker placement is described in detail in Chapter 3. Due to the possible presence of high frequency noise in the digitised position of each marker, data were filtered using a Quintic Spline incorporated within the motion analysis system (Peak Performance Default Optimal Smoothing).

Each female performed one, two second static trial with markers attached, and both feet placed on the force platform. This procedure was standardised; participants were requested to look straight ahead with eyes focussed on the wall at eye level to reduce any unnatural shift in body weight. The static trial was consistently performed in the control (neutral) shoe condition. Data collected during the two second interval provided standing joint angles for adjustment of anatomical angles during running (Hunt et al., 2006).

5.2.4.2. Gait analysis.

The gait experiments were performed with each female running along a 10 meter runway, through the capture volume, and striking the force platform with the right foot. Running velocity was monitored at 3.5 $m.s^{-1}$ ± 5% (a speed determined in Pilot Study of Study One) using two, one meter high infrared timing gates located along the 10-m concrete runway, 2.5 meters either side of the centre of the force plate. Participants were permitted practice trials to ensure the step landed in the middle of the force platform, and the gait was not altered in order to land on target. Trials that did not land on the force platform, or

that were not within the required range of velocity were discarded. A successful trial was one which had the right foot land on the centre of the platform, with all eleven markers visible throughout the stance phase of running gait. Ten successful trials were then performed in each condition. Test conditions were selected randomly with the order undisclosed to the participant.

5.2.5. Three-Dimensional data analysis.

Kinematic data were calculated during the stance phase of running gait, and peak values were obtained for lower limb movement in the sagittal, transverse and frontal plane of motion. Each variable of interest was isolated from the data, and ten trial means for each participant in each condition were used to obtain group means. Force and moment data were presented relative to body weight. When calculating the external joint moments, each limb segment (thigh, shank, foot) was idealised as a rigid body and the positions of the joint centres were located relative to the position of the skin markers. Knee and ankle joint centres were therefore located based on the markers positioned at the lateral joint line of the knee, and the lateral malleolus respectively. The knee external adductor moment was then calculated using a customised MATLAB code. Inverse dynamics techniques were used, with data from marker position, force plate and inertia parameters provided by Dempster (Dempster et al., 1955). Peak moment was calculated as the maximum during the stance phase of gait, and average peak moment values over 10 trials were presented for each participant and shoe condition.

Moment arm length of the ground reaction force about the knee joint centre was calculated for each trial performed by every participant in all four footwear conditions. The length was calculated with the same method as is discussed in the previous chapter (Equation 4.5). Peak moment arm was calculated as that which occurred at the time of peak moment during stance, and the occurrence time was noted. A repeated measures ANOVA was used to assess the difference in moment arm length among the mature females between the neutral, medial, lateral and orthotic footwear conditions. A

regression test was then performed to assess the relationship between maximum moment arm length and peak knee external adductor moment.

Effect size for change in biomechanical variable for each shoe condition was calculated. The size of the effect was determined from the values presented by Cohen (1988), whereby an effect size of 0.01 was small; 0.059 was medium; and 0.138 was large. Significance between footwear conditions was determined with a repeated measures ANOVA conducted with participant means. Data was significant at the $p < 0.05$ level. Where results were significant, a post hoc Tukey Honeslty Significant Differences (HSD) test was performed to identify where specific statistical differences existed between footwear conditions (Equation 5.1).

$$HSD = q(k,df_E) \text{ sqrt } MSE/n$$

equation 5.1

Where HSD is the honestly significant difference, q is the studentised range distribution for k (the number of groups) and df_E (degrees of freedom related to the error term). MSE is the mean square error from the ANOVA, and n is the sample size.

5.3. Results.

5.3.1. KOOS Scores and Somnio Line-Up assessment outcomes.

5.3.1.1. KOOS Results.

The knee osteoarthritis survey (Appendix B) was completed by each mature female prior to the initiation of the gait assessment. This survey assessed each individual's opinion about the condition of their own knee, and associated conditions. Group results are presented in Table 5.1. Average results were initially isolated for each subscale, and an outcome profile was plotted (Figure 5.6). Results indicated all females as free from symptoms of knee osteoarthritis and injuries (mean 89.2, sd 2.6). However, a slight discrepancy was illustrated between the subscales, with both quality of life and sports and recreation subscales displaying lower scores than the other three categories.

Table 5.1. Participant information table including 10 km run times and KOOS scores.

MATURE FEMALE RUNNERS		
AGE (YEARS)		**50.55 (4.5)**
MASS (KG)		**58.33 (5.6)**
10-KM TIME (MINS)		**60.9 ± 2.96**
KOOS SCORE		**89.2**
	PAIN	89.7 (2.6)
	SYMPTOMS	89.1 (3.4)
	ACTIVITIES OF DAILY LIVING	93.2 (1.9)
	SPORT AND RECREATION	87.4 (2.4)
	QUALITY OF LIFE	86.6 (2.9)

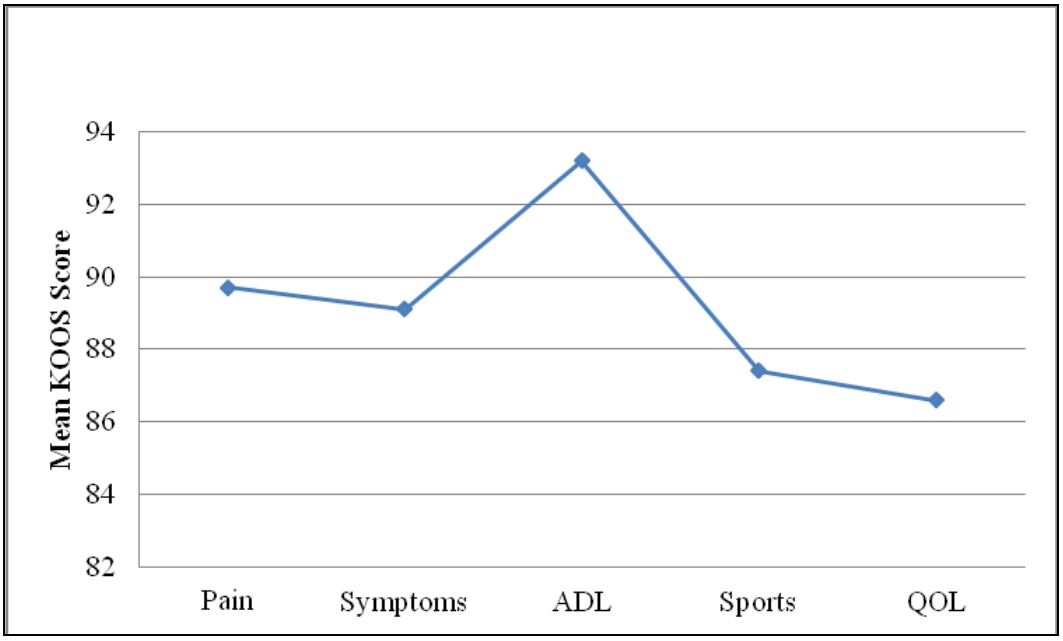


Figure 5.6. Line graph illustrating average KOOS results from the mature female runners.

A correlation test between KOOS results and age is performed and discussed in Chapter 6. However, although all mature females have been categorised as free from symptoms of osteoarthritis or knee injuries in each study, since KOOS results were also available for the earlier studies of this thesis, a retrospective comparison was made between the scores obtained over the three studies (an approximate three year period) (Figure 5.7). In addition, since six of the participants were common to all three studies of this thesis, a comparison was made between scores elicited over a three year period for these individuals (Figure 5.8). Although no significant difference was demonstrated in the results, Figure 5.7 illustrates a variation in results over the three years. The KOOS scores produced by the six constant participants are displayed in Figure 5.8. In general, the graphs in Figures 5.7 and 5.8 suggest that the symptoms of osteoarthritis and knee injury appear to progress over the three year period, with four from five subscales eliciting lower scores in the third year compared to the first (indicative of worse symptoms). Similarly, in all three assessments, the lowest score was seen in the quality of life

subscale. However, results from an ANOVA test showed no difference in score was significant (p = 0.24).

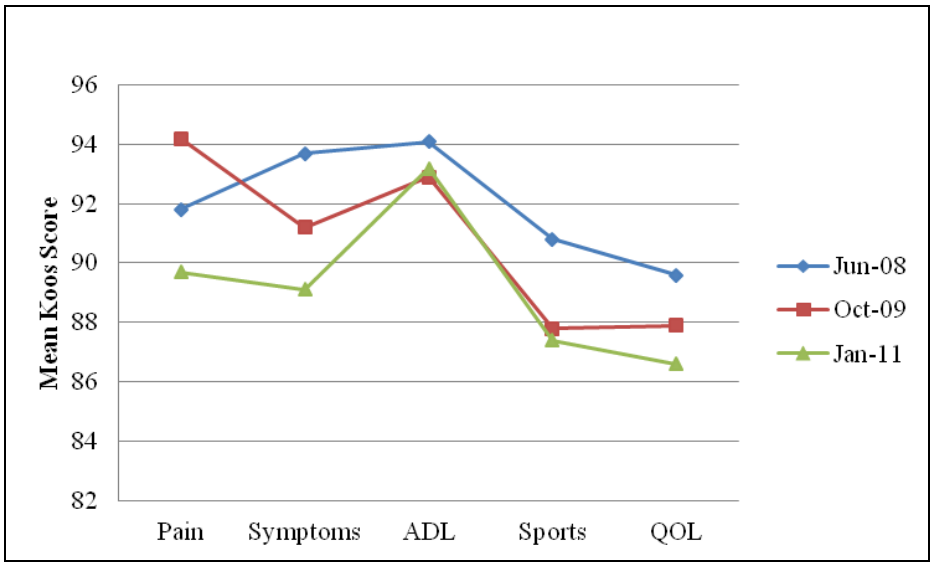


Figure 5.7. Line graph illustrating KOOS scores from mature female runners over three years.

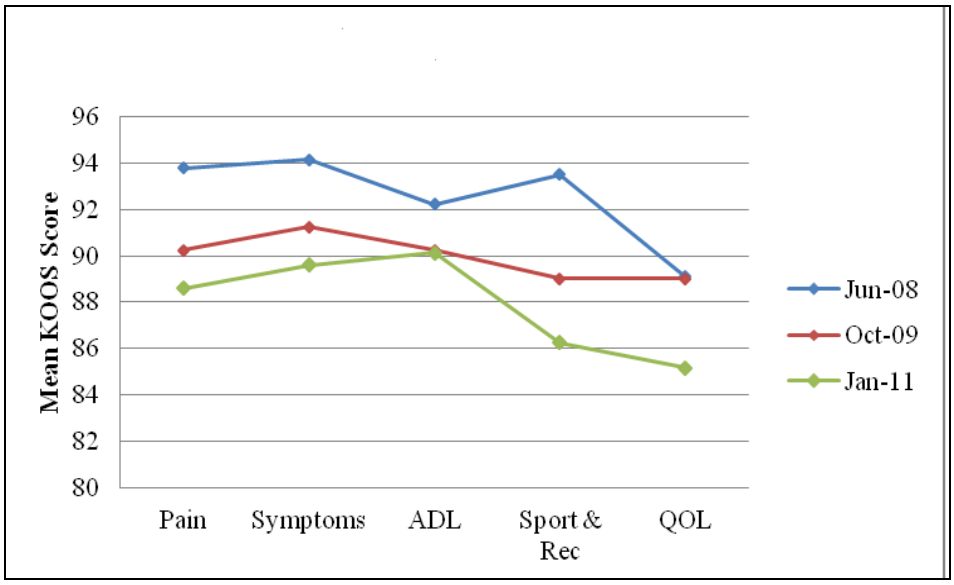


Figure 5.8. Line graph illustrating KOOS scores for the consistent 6 mature female runners over three years.

5.3.1.2. Somnio Line Up Device results.

Results from the Somnio line up device and archometer assessment showed all females to have moderate arch heights. The outcome suggested all females were mild pronators. The orange (medium) foot bed was added to the shoe to act as the neutral condition in the assessment.

5.3.2. Kinematic results during stance phase of running gait.

A numerical comparison of gait variables with changes in footwear condition is presented in Table 5.2. Data include mean peak values of the gait variable, alongside occurrence times, and peak angular velocity, presented in radians per second. Results from the repeated measures ANOVA are presented with each variable, and significant differences between footwear conditions are categorised at the $p<0.05$ alpha level. A table including individual subject means for each peak variable is presented in Appendix F.

Table 5.2. Summary of kinematic values produced by 20 mature female runners.

GAIT VARIABLE (PEAK VALUE)			SHOE CONDITION				SIGNIFICANCE LEVEL
			NEUTRAL	MEDIAL	LATERAL	ORTHOTIC	
ANKLE/ SUBTALAR JOINT	Rearfoot Eversion	Peak Angle (degrees)	14.03 (±3.7)	13.65 (±4.9)	15.69 (±4.68)	8.47* (±4.2)	P = 0.00003*
		Occurrence time (seconds)	0.096 (±0.005)	0.098 (±0.001)	0.10 (±0.06)	0.098 (±0.005)	P = 0.49
		Eversion velocity (rad.s^{-1})	4.03 (±0.94)	4.15 (±0.91)	4.20 (±0.73)	3.79 (±0.39)	P = 0.32
	Foot Abduction	Peak Angle (degrees)	11.7 (±5.18)	11.95 (±9.02)	11.83 (±7.97)	11.2 (±7.69)	P = 0.99
		Occurrence time (seconds)	0.16 (±0.04)	0.15 (±0.03)	0.16 (±0.05)	0.15 (0.01)	P = 0.18
		Abduction velocity (rad.s^{-1})	2.24 (±1.0)	1.86 (±0.6)	1.86 (±0.6)	2.10 (±1.1)	P = 0.44
	Dorsiflexion	Peak Angle (degrees)	19.9 (±3.64)	18.2 (±3.54)	18.8 (±3.88)	17.7 (±1.83)	P = 0.24
		Occurrence time (seconds)	0.13 (±0.02)	0.13 (±0.03)	0.13 (±0.04)	0.13 (±0.03)	P = 0.95
		Dorsiflexion velocity (rad.s^{-1})	10.16 (±0.4)	10.52 (±0.5)	10.34 (±0.6)	10.5 (±0.5)	P = 0.37
KNEE JOINT	Knee Flexion	Peak Angle (degrees)	35.25 (±5.58)	33.54 (±5.14)	32.58 (±5.66)	34.38 (±6.77)	P = 0.51
		Occurrence time (seconds)	0.17 (±0.05)	0.18 (±0.02)	0.18 (0.03)	0.18 (0.02)	P = 0.15
		Flexion velocity (rad.s^{-1})	15.00 (±0.5)	14.6 (±0.4)	14.9 (±0.3)	14.7 (0.4)	P = 0.10
	Knee Abduction	Peak Angle (degrees)	13.13 (±7.37)	12.31 (±4.91)	9.69 (±5.65)	9.74 (±5.72)	P = 0.17
		Occurrence time (seconds)	0.21 (0.0002)	0.23 (±0.0004)	0.22 (±0.0003)	0.23 (±0.0009)	P = 0.02*
		Abduction velocity (rad.s^{-1})	1.77 (±0.2)	1.72 (±0.2)	1.82 (±0.2)	1.69 (±1.2)	P = 0.77
	Knee Internal Rotation	Peak Angle (degrees)	-14.4 (±2.47)	-12.99 (±2.84)	-14.13 (±3.06)	-9.81* (±2.68)	P = 0.00021*
		Occurrence time (seconds)	0.12 (±0.02)	0.17 (±0.02)	0.12 (±0.09)	0.17 (±0.001)	P = 0.99
		Rotation velocity (rad.s^{-1})	2.65 (±0.2)	2.40 (0.2)	2.50 (0.1)	2.22* (±0.2)	P = 0.011*

*Differences were statistically significant compared to neutral condition at $p<0.05$ level.

Table 5.3 indicate results from a regression analysis performed between each variable and age. Specific kinematic data are discussed in detail in the following tables, and where appropriate, data are presented in graphical form to illustrate trends and magnitudes of differences in study variables with shoe conditions.

Table 5.3. Statistical output illustrating results from regression analyses between each variable and age.

STATISTICAL ANALYSIS OUTPUT	ANKLE/SUBTALAR JOINT			KNEE JOINT		
	AGE:RF EV	AGE:DF	AGE:FABD	AGE:KFLEX	AGE:KABD	AGE:KINT
Correlation Coefficient	0.62	0.096	0.50	0.06	0.30	0.23
R Square	0.39	0.0094	0.25	0.004	0.088	0.05
Observations	20	20	20	20	20	20
P-Value (2 sf)	0.0033*	0.68	0.025*	0.79	0.20	0.32

*Regression was statistically significant at $p<0.05$ level.

5.3.2.1. Ankle/Subtalar joint.

5.3.2.1.1. Rearfoot eversion.

Comparisons of the mean peak rearfoot eversion angles produced by each mature female indicated a significant difference between conditions ($p<0.05$). Results from the post-hoc analysis (Tukey's Honestly Significant Difference test) indicated that the significance lay between the orthotic intervention and the other three footwear conditions. The orthotic condition produced an average decrease of 5.56 degrees in peak rearfoot eversion when compared to the neutral condition. This difference is highlighted in Figure 5.9, which graphically illustrates the differences between conditions. Here is it shown that the orthotic condition produced the lowest rearfoot eversion value. The lateral footwear condition increased rearfoot eversion by 1.66 degrees above that produced in the neutral condition, although this increase was not significant ($p>0.05$, effect size 0.046). A sample graph illustrating rearfoot eversion angle throughout the stance phase of gait, and the discrepancy between the neutral and orthotic condition is illustrated in Figure 5.10. This graph displays mean data from Subject 10. There was no significant difference in the occurrence times between conditions ($p>0.05$), and results from a repeated measures ANOVA with the mean rearfoot angular velocity produced by each participant illustrated no significant differences between conditions. The relationship between rearfoot eversion and age within the group of mature females is illustrated in Figure 5.11. Results from the regression analysis indicate a significant positive linear relationship ($r = 0.62$, $p<0.05$), suggesting an increase in rearfoot eversion occurred across an age gap of 17 years (Table 5.3).

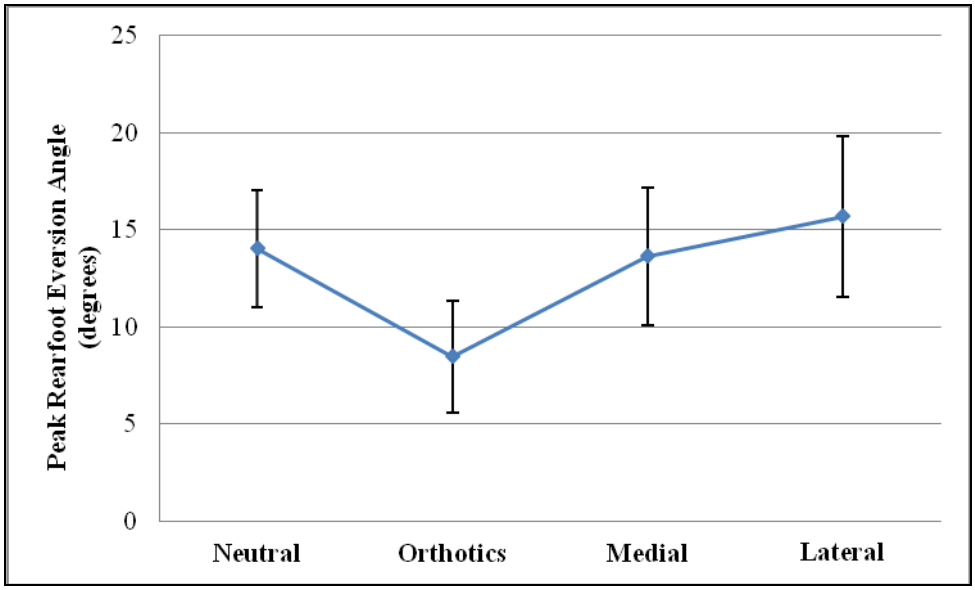


Figure 5.9. Group mean rear foot angles with changes in footwear condition.

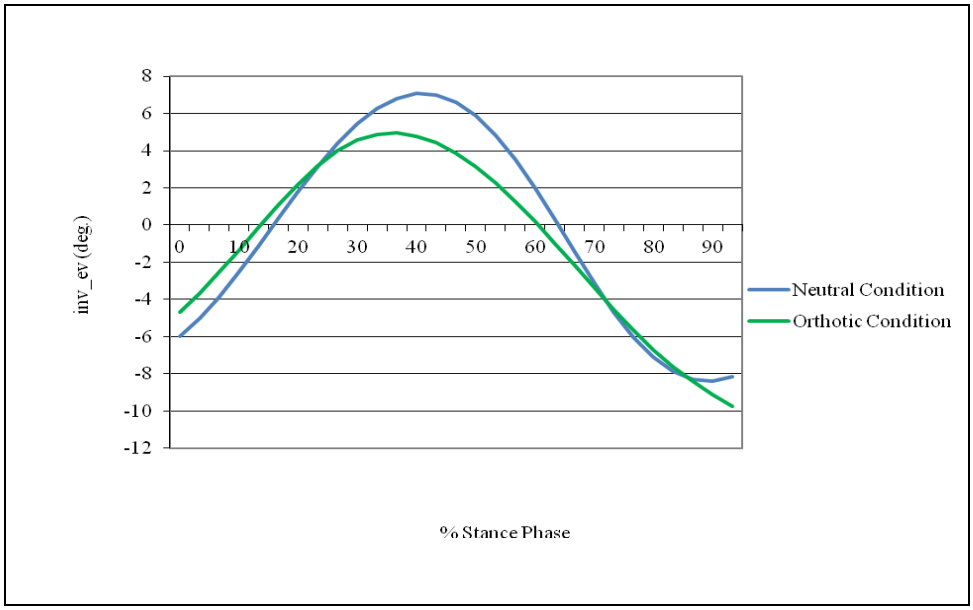


Figure 5.10. Mean rearfoot eversion angle data for neutral and orthotic footwear conditions. Subject 10.

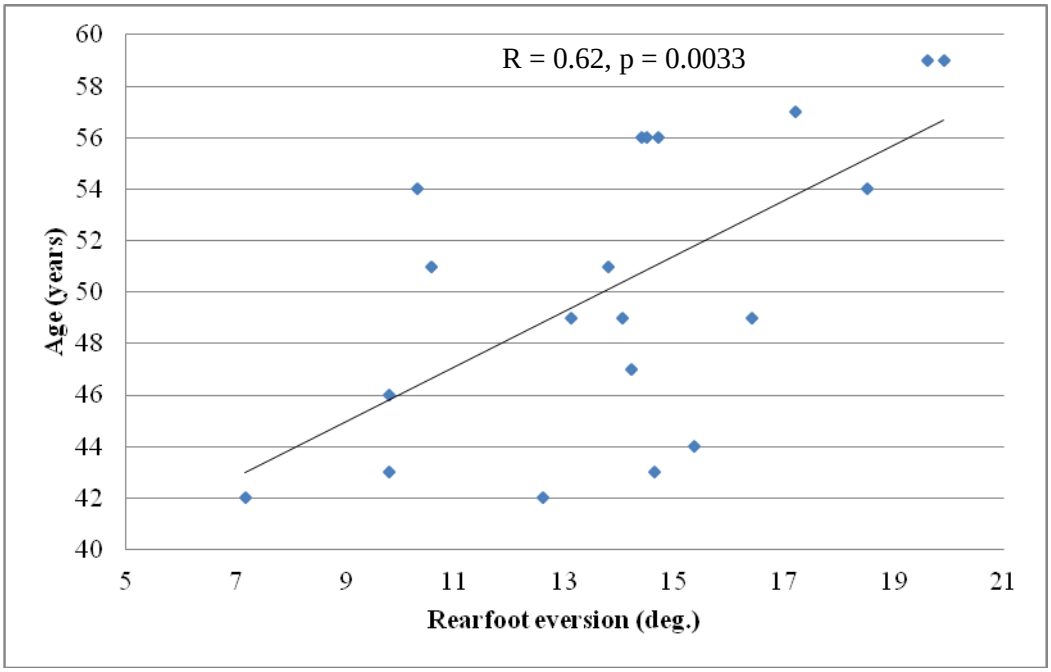


Figure 5.11. Scatter diagram illustrating linear relationship between peak rearfoot eversion angles and age among mature female runners.

5.3.2.1.2. Foot abduction.

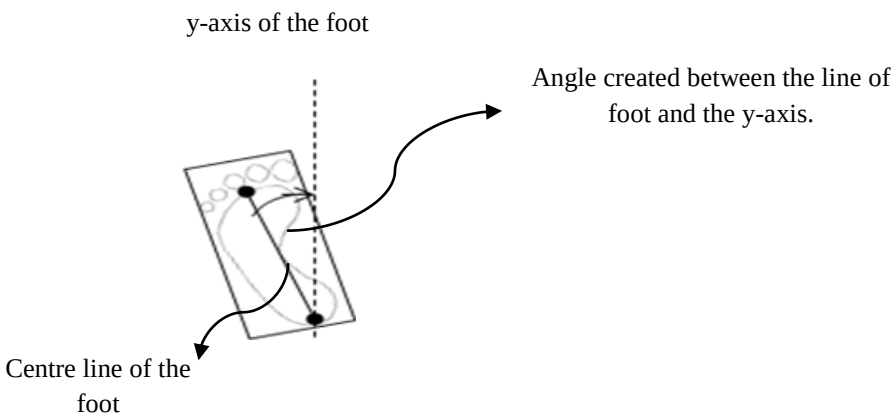


Figure 5.12. Annotated diagram of the foot abduction angle calculated. Source (foot): Microsoft Office Clip Art 2011.

Mean foot abduction data produced by the mature females is presented in Table 5.2. Among the mature female runners, results from this variable were consistent between conditions, with no significant differences identified between footwear ($p>0.05$). Despite this, it was seen that the medial wedge produced the highest value (11.95^{o}), compared to the orthotic condition which produced the lowest (11.2^{o}) although the values varied by less than one degree (Table 5.2). The standard deviation of the means produced by the neutral condition was noted to be smaller than those produced from the three footwear interventions.

Mean occurrence of peak foot abduction was not significantly different between footwear conditions ($p>0.05$). Similarly, peak angular velocity produced non significant differences, although the highest foot abduction velocity was illustrated in the neutral condition when compared to the three footwear interventions ($p>0.05$). Results from the regression analysis indicated a significant positive correlation between foot abduction and age ($r = 0.50$, $p<0.05$), suggesting that the mature females at the higher end of the age scale ran with increased abduction of the forefoot during stance. This relationship is highlighted graphically in Figure 5.13.

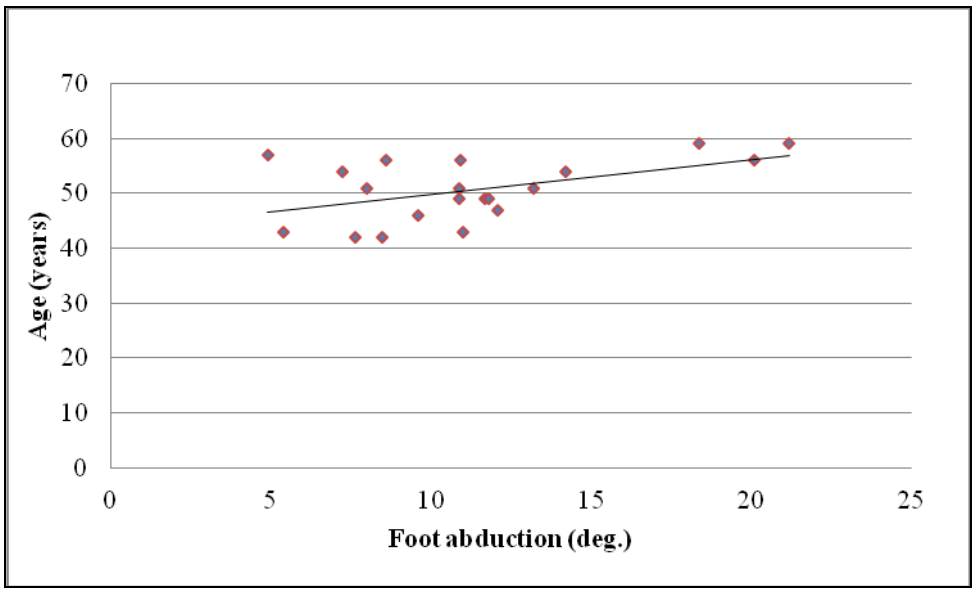


Figure 5.13. Scatter diagram illustrating positive relationship between foot abduction and age.

5.3.2.1.3. Ankle dorsiflexion.

Peak ankle dorsiflexion occurred during the later stage of mid-stance phase, at approximately 50 % of stance (Figure 5.14). Within the neutral condition, mature females produced an average dorsiflexion angle of 19.9^{o} (±3.64^{o}); the highest peak angle compared to the three footwear interventions. The lowest peak angle was shown in the orthotic condition (17.7^{o}). However, no significant differences were observed between the four footwear conditions ($p>0.05$, effect size 0.084) (Table 5.2). Peak dorsiflexion angle was consistently shown to act at 0.13 seconds, with no significant difference seen between footwear conditions ($p>0.05$). Peak dorsiflexion velocity was lowest in the neutral condition compared with the orthotic, medial and lateral wedge, although again this difference was not significant ($p>0.05$). Among the 20 mature females, peak dorsiflexion angle did not differ significantly with age ($r = 0.22$, $p>0.05$).

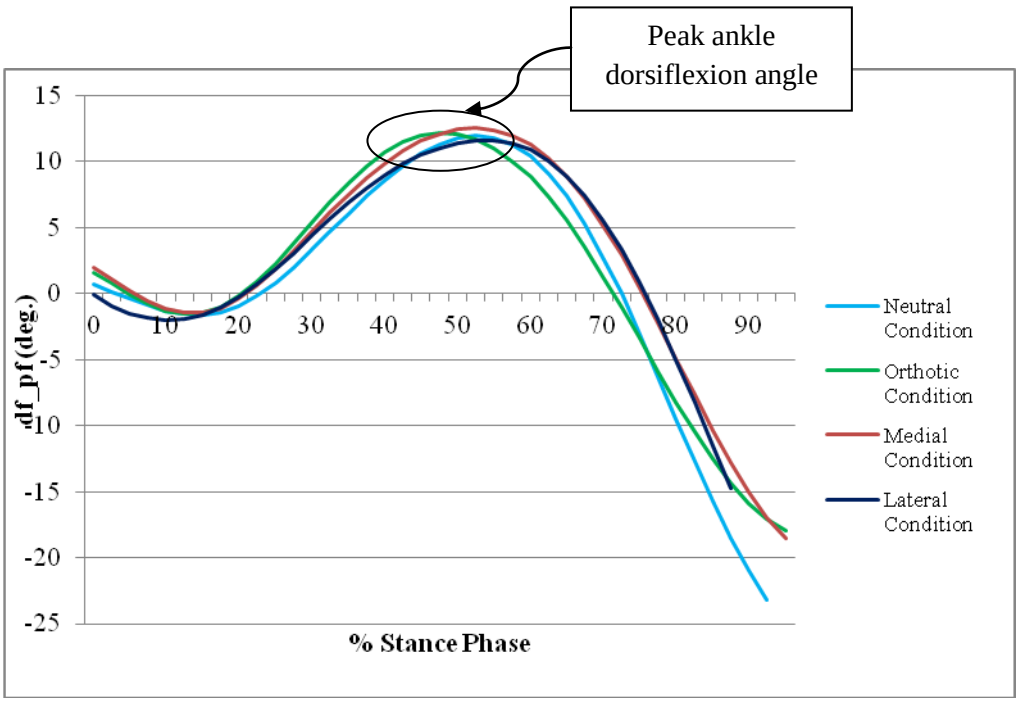


Figure 5.14. Mean peak dorsiflexion angle time history produced by subject 7 in all four conditions.

5.3.2.2. Knee joint.

5.3.2.2.1. Knee flexion.

The peak knee flexion angles, occurrence times, and peak angular velocities produced by the mature females were not significantly different between footwear conditions ($p>0.05$) (Table 5.2). Similarly, there was no significant correlation between knee flexion angle and age ($r = 0.06$, $p>0.05$). The largest variation in peak angle (2.67^{o}) for this variable was demonstrated between the neutral shoe, and the lateral wedge condition (Figure 5.15). A sample knee flexion time history is shown in Figure 5.16, illustrating mean data produced by Subject 12. The graph shows the knee moving from a low degree (approximately 5°) of flexion at heel strike, to maximum flexion at mid stance, where flexion is calculated as a negative value.

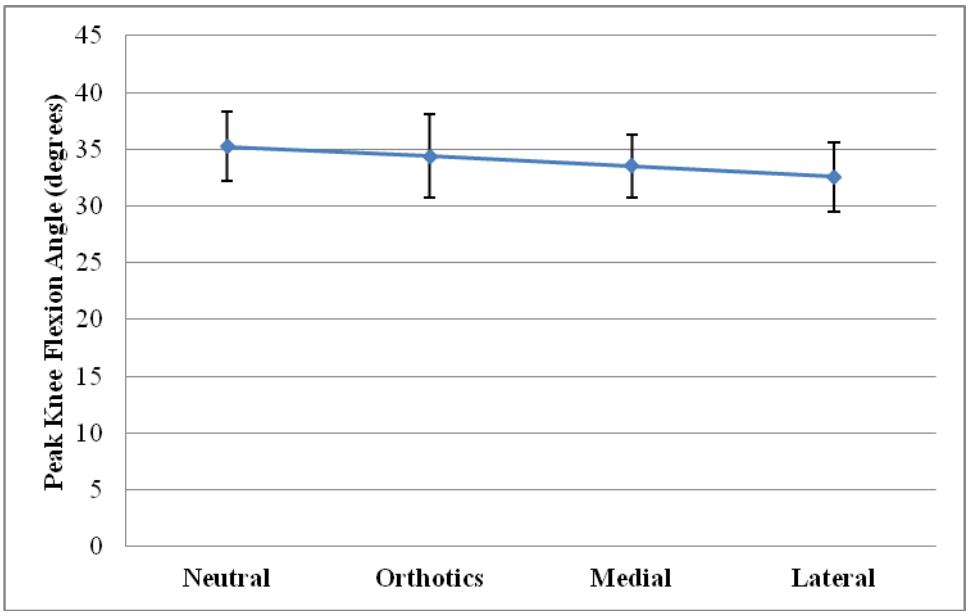


Figure 5.15. Line graph illustrating mean peak knee flexion angle produced by the mature female runners in all four conditions.

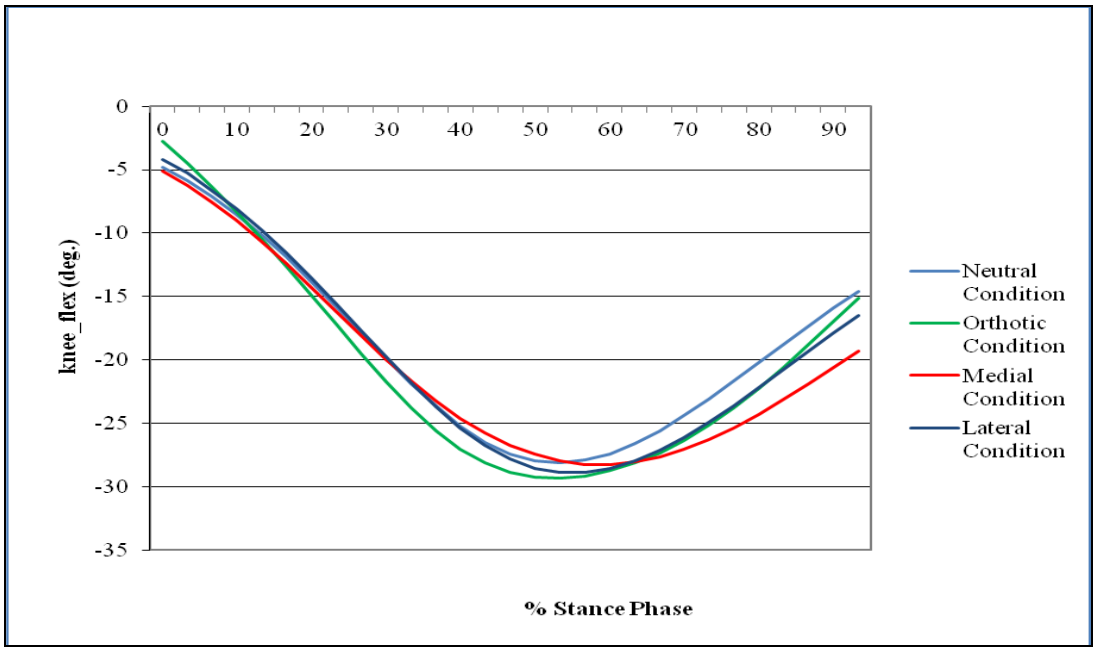


Figure 5.16. Knee flexion graph illustrating mean data for Subject 12.

5.3.2.2.2. Knee abduction (varus).

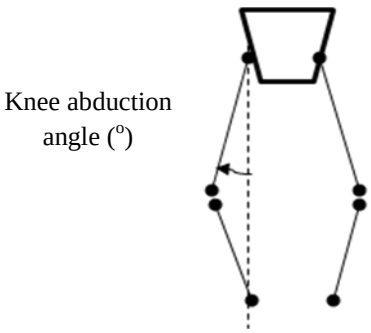


Figure 5.17. Diagram illustrating the knee abduction angle calculated.

Peak knee abduction angle, calculated as the movement of the knee joint from the midline (Figure 5.17), showed no significant differences between conditions ($p>0.05$), and no significant correlation was seen with age ($r = 0.30$, $p>0.05$). Although not significant, small differences were shown between footwear conditions, with the orthotic intervention (effect size 0.064) and the lateral wedge (effect size 0.067) both producing a decrease in the angle compared with the neutral and medial conditions (Table 5.2). Figure 5.18 illustrates a line graph of group mean results. A sample abduction angle graph illustrating mean data for the neutral and orthotic condition is displayed in Figure 5.19. A significant difference was however seen in the mean occurrence time of peak knee abduction ($p<0.05$). Results from the post hoc analysis indicated that this significant difference lay between occurrence time in the neutral condition and the orthotic condition ($p<0.05$), indicating that the neutral shoe resulted in a peak angle significantly earlier than the orthotic conditions. Despite this difference, no significant difference was seen in the velocity of knee abduction between footwear conditions ($p>0.05$). Angular velocity of knee abduction did produce the lowest results in the orthotic condition; however results from the repeated measures ANOVA indicated this difference as non-significant ($p>0.05$).

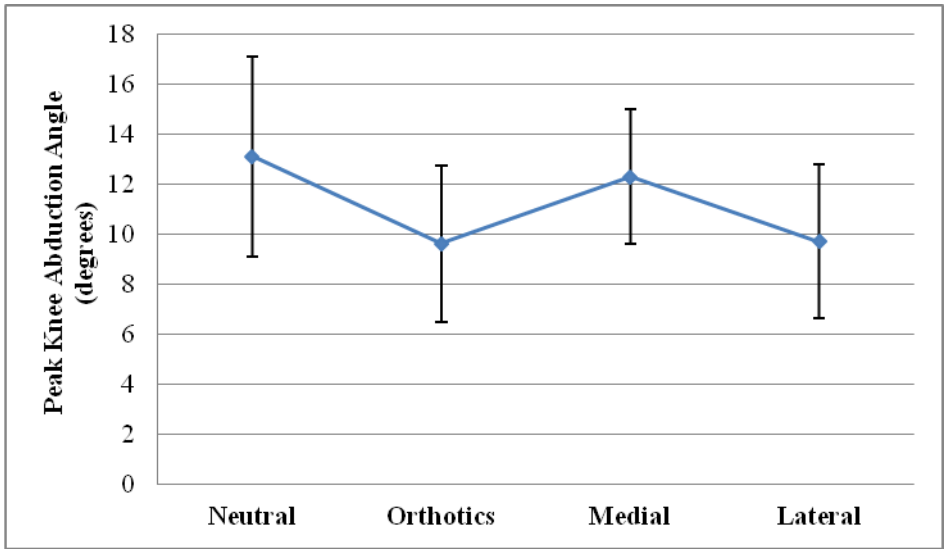


Figure 5.18. Line graph illustrating difference in peak knee abduction between the four footwear conditions.

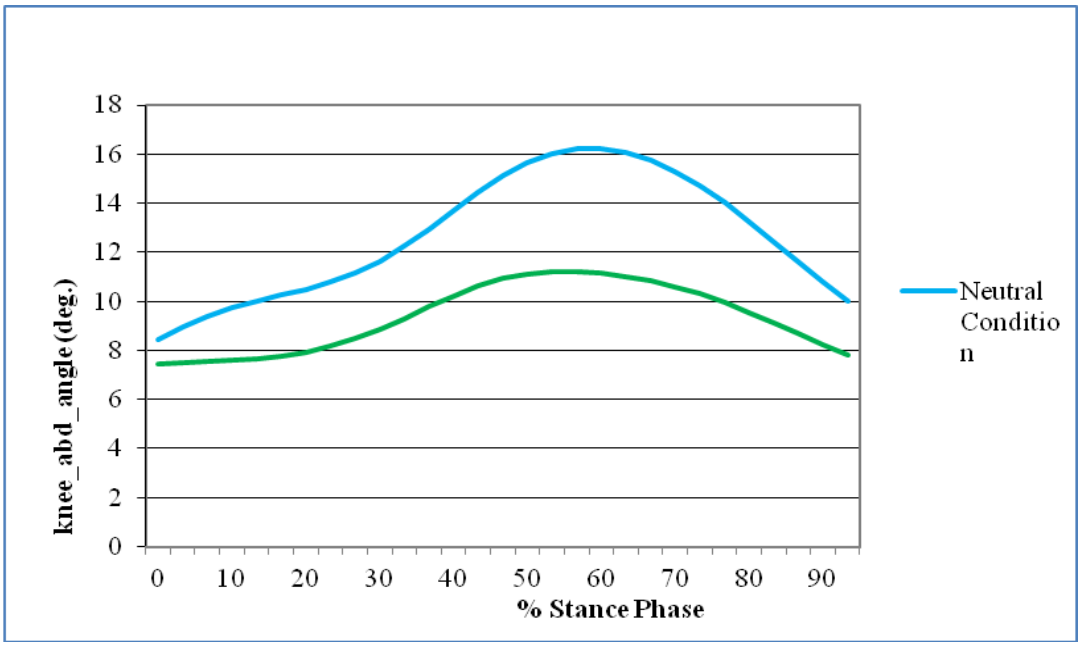


Figure 5.19. Knee abduction angle line graph produced by Subject 20 in the neutral and orthotic conditions.

5.3.2.2.3. Knee internal rotation.

Results produced from the mature females indicated an average internal rotation value of 12.6^{o} across all four conditions. Results from a repeated measures ANOVA indicated a significant difference in knee internal rotation values for the four footwear conditions, with the orthotic conditions producing the lowest value of rotation ($p<0.05$) (Figure 5.20). The post hoc analysis indicated that the orthotic condition produced significantly lower values of rotation when compared with both the neutral and the lateral footwear conditions ($p = 0.0000017$; $p = 0.000022$). Figure 5.21 visually displays this difference,

with mean data from Subject 5 clearly showing reduced knee internal rotation values with the orthotic intervention.

The mean occurrence time of peak knee internal rotation was not significantly different between footwear conditions. However, velocity of internal rotation elicited a significant difference between conditions ($p<0.05$). Post hoc analysis showed the significant difference to lie between the neutral and orthotic condition, with the neutral shoe producing an angular velocity 0.43 rad.s^{-1} faster than the orthotic intervention. Results from a regression analysis indicated no significant correlation between knee internal rotation and age ($r = 0.23$, $p = 0.3$) (Table 5.3).

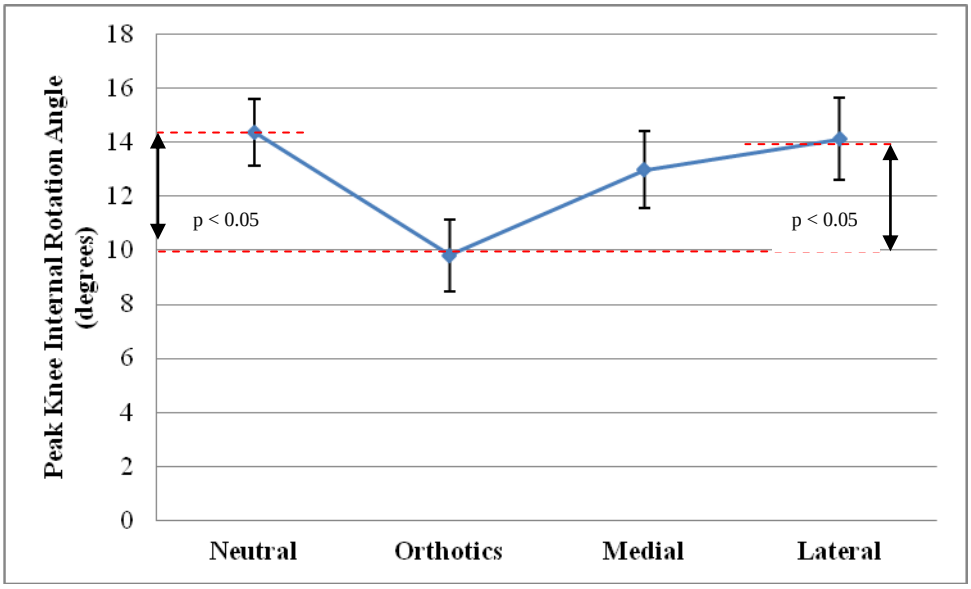


Figure 5.20. Line graph illustrating a significant difference between peak knee internal rotation between the four footwear conditions.

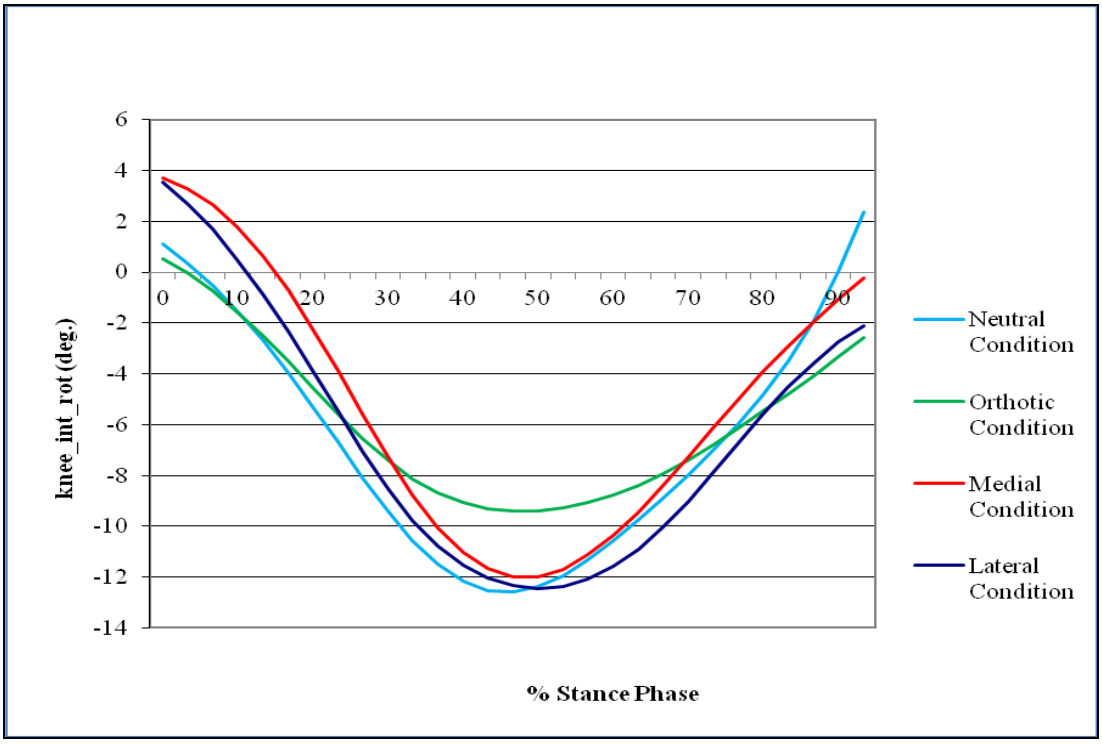


Figure 5.21. Knee internal rotation graph representing mean data for the four footwear conditions (Subject 5).

5.3.3. Knee Moments: the external knee adductor moment and moment arm.

The knee external adductor moment was calculated throughout the stance phase of running gait, for each trial performed by every mature female. A sample external adductor time history is displayed in Figure 5.22, highlighting the second, larger peak as the peak used for the averages. Moment arm length was calculated at the time of peak moment. A summary of individual means for both variables in each footwear condition, and average occurrence time are presented in the Appendix (Appendix F). Moment data are presented in both raw form (Nm) and scaled to body mass (Nm.kg). Group means for both peak moment and moment arm, and average occurrence times are presented in Table 5.4.

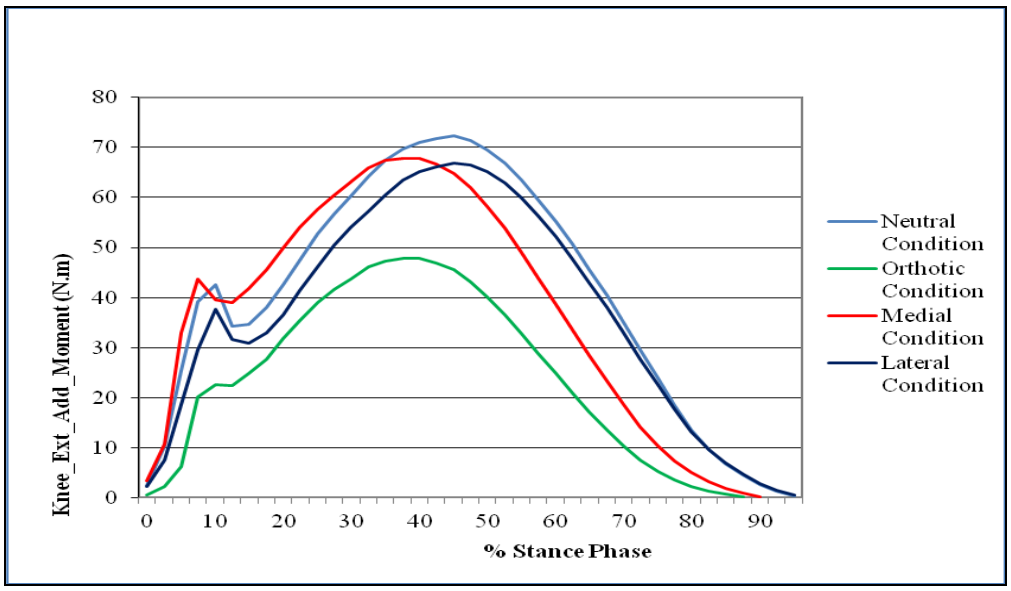


Figure 5.22. Sample knee external adductor moment graph, illustrating the peak external adductor moment. Data taken from Subject 15.

Table 5.4. Summary of group means and occurrence time for peak knee external adductor moment (Nm/kg) and moment arm length (m).

KNEE EXTERNAL ADDUCTOR MOMENT		FOOTWEAR CONDITION				SIGNIFICANCE LEVEL (P < 0.05)
		NEUTRAL	ORTHOTICS	MEDIAL	LATERAL	
Peak Moment (Nm/kg)	Peak Angle (degrees)	1.15 (±0.12)	0.92 (±0.19)	1.05 (±0.17)	1.02 (±0.015)	P = 0.000538*
Peak Moment arm (cm	Peak Length (metres)	-0.020 (±0.001)	-0.019 (±0.001)	0.020 (±0.003)	0.018 (±0.003)	P = 0.07*
Occurrence time (seconds)		0.20 (±0.016)	0.20 (±0.013)	0.20 (±0.0016)	0.19 (±0.017)	P = 0.13

A breakdown of statistical results are presented in Tables 5.5 a and b. Results from a repeated measures ANOVA of peak moment illustrated a significant difference in the peak knee external adductor moment with the different footwear conditions. Post hoc analysis showed this difference to lie between the neutral and orthotic condition ($p < 0.05$), the neutral and the lateral condition ($p < 0.05$), and the medial and the orthotic condition ($p < 0.05$) (Table 5.5a).

Similarly, statistical results from the moment arm data showed a significant difference in this variable between footwear conditions. As shown in Table 5.5b, this difference lay between the neutral and the orthotic condition ($p < 0.05$), and the neutral and the lateral condition ($p < 0.05$)

Table 5.5 a and b. Statistical analysis of difference in knee external adductor moment (a) and moment arm length (b) between footwear conditions.

(a)

Footwear Condition	Neutral	Orthotic	Medial	Lateral
Neutral	—			
Orthotic	**0.00076***	—		
Medial	0.061	**0.024***	—	
Lateral	**0.0094***	0.065	0.57	—

(b)

Footwear Condition	Neutral	Orthotic	Medial	Lateral
Neutral	—			
Orthotic	**0.0015***	—		
Medial	0.51	0.16	—	
Lateral	**0.025***	0.94	0.22	—

5.3.3.1. Regression analysis between moment and moment arm length.

Results from a regression analysis between peak knee external adductor moment (KEAM) and moment arm length for each footwear condition are presented in Table 5.6. The correlations are presented graphically in Figure 5.23.

Table 5.6. Results from a regression analysis of peak moment and moment arm length.

STATISTICAL ANALYSIS OUTPUT	KEAM: Moment arm length.			
	Neutral	Orthotics	Medial	Lateral
Correlation Coefficient (R)	0.65	0.47	0.41	0.63
R Square (R^2)	0.42	0.22	0.17	0.39
Observations	20	20	20	20
P-Value (2 sf)	0.0019*	0.037*	0.071	0.003*

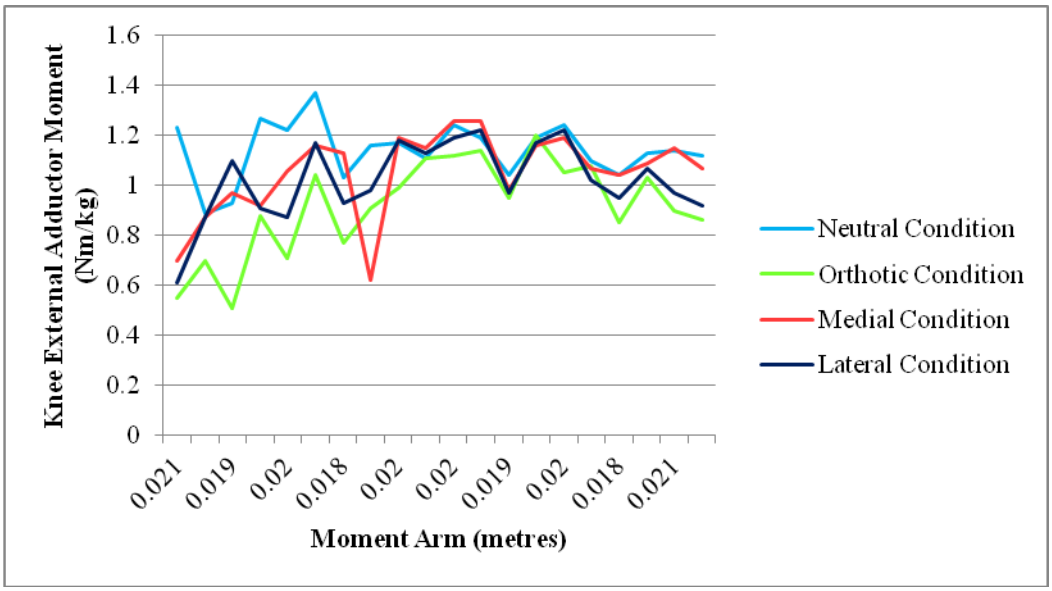


Figure 5.23. Graphical illustration of relationship between peak knee joint moment and moment arm length produced by each female in each footwear condition.

5.4. Discussion.

The purpose of this biomechanical study was to assess the effect of different footwear interventions on the running gait of mature females. Twenty females with a mean age of 50.6 ± 5.6 years were recruited for participation on a volunteer basis. Each female ran along a marked pathway, while the motion was analysed with an eight camera motion capture system, synchronised with a force platform embedded in the ground. The four footwear conditions included a neutral shoe, a medial wedge, a lateral wedge, and an orthotic intervention. The Somnio 6 mm medial and lateral wedges were composed from rubber-plastic, and inserted into the shoe under the medium foot bed. The 6 mm wedges began at the heel, and continued to the start of the medial arch (medial wedge) and the lateral mid foot (lateral wedge), reducing to the standard height at the forefoot. The purpose of the medial wedge was to restrict subtalar joint pronation. Conversely, the purpose of the lateral wedge was to reduce the lateral tilt of the foot and subsequent abduction of the lower leg and knee joint. The orthotic intervention incorporated both mechanisms, although rearfoot eversion control was achieved through increased medial support at the heel and under the medial longitudinal arch, restricting rearfoot eversion shortly after heel strike. The lateral wedge was focussed along the length of the lateral sole and into the forefoot, aiming to restrict lateral movement of the foot and lower leg at mid stance.

Although several variables of gait were not altered with the different footwear conditions, significant group trends observed in specific aspects of human movement were identified. The use of post hoc analyses identified where these differences lay, and the overall effect of each footwear condition on the running gait of mature females was assessed.

5.4.1. Initial Assessments: KOOS and Somnio Line-Up Device.

5.4.1.1. KOOS outcome.

All female participants were within the 40 to 60 year age bracket, and as such were proposed to be at high risk for osteoarthritis of the knee joint (Guccione, 1997). Additionally, as each female was known to undertake a session of physical activity a minimum of three times per week, the link between physical activity participation and susceptibility to injury or knee OA could be considered to heightens the position of this group to a high risk category (Imeokparia, Barrett, Arrieta, Leaverton, Wilson Hall and Marlowe, 1994). Based on this, the results from the KOOS assessment were unexpected, with all females categorised as free from symptoms and signs of osteoarthritis.

The results from the survey showed the lowest score, indicative of increased signs of osteoarthritis, to be present in the quality of life subscale. Comparatively, the subscale showing the least signs of osteoarthritis was the activities of daily living. These results suggest that although categorised as mild, all the mature females within this investigation suffer an element of discomfort that may have an impact on their lives.

It has been suggested that there is a poor association between symptomatic and radiographic levels of osteoarthritis (Dawson, Fitzpatrick, Fletcher, & Wilson, 2004). Thus, despite the high scores from the KOOS assessment, it is possible that the study participants have a degree of radiographic osteoarthritis that does not yet elicit symptoms. According to Dawson and colleagues, the extent of radiographic osteoarthritis at the knee joint does not predict the likely progression to a symptomatic state (Dawson, Fitzpatrick, Fletcher, & Wilson, 2004). However, the presence of osteoarthritis among this specific group of mature females is currently unknown, and will be assessed in the magnetic resonance imaging study, Chapter 6. Here a direct correlation with KOOS scores will investigate this relationship further.

Figure 5.7 illustrates the average KOOS scores produced by a group of mature females over the past three years. Although the group did not consistently contain the same females, six participants were involved in all three studies. The results displayed in the

line graph suggest a general tendency for scores to drop over the three years, although this difference was not significant.

For the six participants that were involved in all three studies, a second comparison of scores across the years occurred (Figure 5.8). Here it was shown that an obvious increase in signs and symptoms of osteoarthritis was observed over the three years, with a difference in score of up to 7.27 seen in the sport and recreation subscale. This suggests that as age of mature females increases, perception of the condition of the knee deteriorates, having the greatest effect during times of physical activity. Although still categorised as symptom free on the KOOS scale, these results support the suggestion that knee condition deteriorates with age, and risk of osteoarthritis increases (Hurwitz et al., 2000).

5.4.1.2. Arch height and pronation: Somnio Line up device.

The Somnio line up device is designed to enable a quick assessment of arch height and pronation level based on position of the foot and movement of the knee joint centre during knee flexion. This procedure was undertaken for each mature female prior to the gait analysis assessment, and the foot bed was selected based on the results. All females were categorised as having a moderate arch height with mild pronation, suiting them to the medium foot bed. As it has been shown that pronation (rearfoot eversion) angles are higher than average among the mature females (Study1 and Study2), the use of the Somnio assessment as a guideline is supported. However, as all mature females were consistently categorised as mild pronators, the variation in pronation values produced in the biomechanical assessment suggests limitations of the device to specifically prescribe footwear. Currently, no research exists to critique the use of this assessment in footwear prescription. Based on results among the mature females in the current study, it is suggested that although this method may provide an indication of alignment, the results are extracted with a low degree of accuracy. To accurately assess rearfoot motion or subtalar joint pronation, three dimensional motion analysis techniques remain superior.

5.4.2. Biomechanical movement patterns during running.

The four footwear conditions included a neutral foot bed, two 6 mm wedges in the medial (medial wedge) and lateral (lateral wedge) directions, and an orthotic intervention incorporating both medial and lateral support. All subjects performed ten running trials in each condition, and results were collaborated.

5.4.2.1. Selection of medial and laterally inclined wedge height.

The 6° incline of both the medial and lateral wedges were selected based on the results of a pilot study conducted prior to onset of data collection for the main investigation. Results from this preliminary investigation with one mature female illustrated no significant changes to gait with either a 2° or 4° inclined wedge on the medial or lateral sides. The 6° medial wedge however showed a significant reduction in the rearfoot eversion and knee internal rotation angles during the stance phase of gait. Similarly, the 6° lateral wedge caused a significant reduction in the first peak knee external adductor moment calculated.

On observation, and when questioned, the participant involved in the pilot study suggested that no discomfort was experienced with either the 6° medial or lateral wedge when running. As such, this level of inclination was selected for investigation, to significantly alter the rearfoot and knee biomechanics among mature females during running.

5.4.2.2. Medial wedge.

5.4.2.2.1. Ankle/Subtalar joint kinematics.

The running trials performed in the medial wedge produced no significant changes in the biomechanics of running gait among the group of mature females. At the foot-ankle complex, subtalar joint pronation is produced from the combination of three movements; foot abduction, ankle dorsiflexion, and rearfoot eversion, the latter being the most dominant variable in subtalar joint pronation. Within this investigation, it was initially hypothesised that when compared with those produced in the neutral condition, running trials performed in the medial footwear condition would produce significantly lower values of rearfoot eversion. Results from the motion analysis showed the medial wedge to have a non significant effect on the peak rearfoot eversion angle ($p>0.05$). A low effect size of 0.0023 was produced, suggesting that 0.2% of change in rearfoot eversion was influenced by the medial wedge. Similarly, no significant difference was observed in occurrence time of peak rearfoot eversion or the angular velocity with the medial wedge. These results therefore suggest little effect of a 6 mm wedge on rearfoot motion.

Rearfoot motion indicates the amount of subtalar joint pronation that occurs during walking and running gait (Kilmartin, & Wallace, 1994). During normal running gait, the subtalar joint passes from a position of supination at heel strike to pronation at early midstance. This joint movement is represented in the sample rearfoot angle graph produced by Subject 10 (Figure 5.10). In the neutral condition, among the mature females, the rearfoot moved from an average of 6 degrees of inversion (subtalar supination) to an average of 14.03 degrees of rearfoot eversion. According to Kilmartin et al., (1994), ideal values of rearfoot eversion peak at an average of 10 degrees, suggesting the peak produced by the mature runners in this investigation were high. Excessive magnitudes of rearfoot eversion and therefore subtalar joint pronation have been associated with a range of running injuries and the development of degenerative conditions at the knee joint (Taunton, McKenzie, & Clement, 1988).

Although research has commonly supported the use of shoe inserts to control for excessive rearfoot motion, the lack of change in eversion angle among the mature females

is in line with the research by Stacoff and colleagues (Stacoff et al., 2001). In one investigation, rearfoot control was intended to be affected by alterations on the lateral side of the shoe, and rearfoot eversion control aimed through a reduction in height of the lateral heel, causing the rearfoot to invert. However, results showed no effect on rearfoot motion, and suggestions were made that the tibiocalcaneal kinematics of every individual are unique, and will respond to footwear interventions differently (Stacoff et al., 2001). It could therefore be suggested that the 6 mm wedge was not great enough to alter the natural kinematics of the mature female runners.

An abundance of research has shown footwear interventions in the form of rearfoot control to produce positive results. The early research performed by Clarke and colleagues (1983), demonstrated a significant decrease in rearfoot angle with a medial heel flare, although the incline was 15 degrees. Similarly, Russell and co-workers showed a decrease in rearfoot eversion with a wedge of 8 degrees, although the significance of this decrease was not tested (Russell, Gruber, O'Connor, Emmerick., & Hamill, 2009). Although the results from this investigation of mature females appear contrary to results in the literature, it is suggested that the height of the medial wedge is a limiting factor in its ability to control excessive rearfoot motion. Additionally, the lack of effect on rearfoot mechanics among the mature females may indicate that the type of material used in the medial wedge or the degree of rearfoot posting was not sufficient to elicit a significant change in rearfoot motion (Stackhouse et al., 2004).

5.4.2.2.2. Knee joint kinematics and moments.

At the knee joint, when compared to the neutral condition the medial wedge produced no significant difference in joint kinematics or knee external adductor moment among the mature female runners. It was hypothesised that the medial wedge would cause a significant decrease in knee internal rotation, due to the proposed reduction in rearfoot eversion and subsequent tibial internal rotation. The lack of significant difference is in line with the results from Eslami and colleagues (2009) who investigated the effect of

wedged orthoses on the movement of the lower limb, and showed a non significant change in tibial and subsequent knee internal rotation ($p = 0.06$).

Analysis of the combined kinematic and kinetic data enabled calculation of the frontal plane knee moment for each female during each running trial. It was hypothesised that the medial wedge would have little effect on this knee external adductor moment. Results from the biomechanical assessment with the mature females supported this hypothesis, with no significant difference in moment elicited with the medial wedge compared to the neutral condition ($p>0.05$). When compared to the results presented in the literature, the data produced by the mature females appears to support the uncertain relationship between medial support and knee external adductor moment. One study performed by Schmalz and colleagues showed an average increase in knee external adductor moment of 9.26% (0.59Nm/kg) with a medial wedge, although the incline was 14 degrees, and therefore not directly comparable to the mature female's data (Schmalz et al., 2006). Conversely, consistent with the findings of the current study, Nester and co-workers assessed the effect of a 10 degree medial wedge, and showed no change in the knee external adductor moment compared to the neutral condition (Nester et al., 2003). Both of these studies however assessed the knee external adductor moment during walking trials. One study involved healthy subjects performing running trials, and showed an average increase of 4.11 % in peak external adductor moment, however it is noted that the medial wedge inserted into the shoe was 26 mm (Franz, Dicharry, Riley, Jackson, Wilder, & Kerrigan, 2008). It is suggested that the relatively small 6 mm thickness of the medial insert in the current study was not sufficient to influence the knee external adductor moment.

5.4.2.3. Lateral wedge.

The lateral wedge incorporated an increase in wedge height on the lateral side of the shoe, beginning at the heel and reducing at the midfoot. It was hypothesised that the running trials performed in the lateral wedge would produce a significant reduction in knee

abduction angle and knee external adductor moment among the mature females, with a consequential increase in the rearfoot eversion and knee internal rotation angles.

5.4.2.3.1. Ankle-Foot complex.

Results from running trials in the lateral wedge reject hypothesis two, as no significant effect was shown in the peak rearfoot angle, occurrence time, or velocity during the stance phase of running gait. However, the lateral condition produced a small change in the average peak angle; increasing the peak by an average of 1.66 degrees ($P>0.05$, effect size: 0.046). As effect sizes of 0.01 are categorised as small, it could be considered that, although not significant, the influence of the lateral wedge on rearfoot eversion was moderate (Cohen, 1988).

The lateral wedge technology placed in the shoe is designed to reduce excessive varus motion at the knee joint, which can in turn rotate the foot segment into a more everted position (Russell et al., 2011). The non significant effect of the lateral wedge on the rearfoot kinematics of the mature females is similar to the results produced by Russell and colleagues. Here it was shown that among the healthy participants, an 8 degree ethylene vinyl acetate lateral wedge produced no significant change in peak rearfoot angle during the stance phase of walking gait. Despite this, many studies have shown lateral wedges to produce significant increases in rearfoot movement, highlighting the negative effect this intervention may have on the lower limb kinematics and the clinical implication during gait.

Butler and co-workers investigated the effect of laterally wedged foot devices on the gait of osteoarthritis patients, and showed a significant increase in peak rearfoot eversion angle during the stance phase of gait (Butler, Barrios, Royer, & Davis., 2009). However the lateral wedge technology used among the patients involved full length orthotic conditions, as opposed to the wedge worn by the mature females. Similarly, Souza and colleagues demonstrated a 10 degree laterally wedged sandal to produce a significant increase in rearfoot eversion among healthy individuals (Souza, Pinto, Trede, Kirkwood, pretence, & Fonseca, 2009). Here it was suggested that distal factors that increase eversion at the foot-ankle complex may affect lower extremity kinematics. However it is important to note that the lateral wedges in the sandal were only inclined in the forefoot,

and the rearfoot eversion was significantly increased at mid to late stance. No significant difference was identified in the peak rear foot eversion during heel strike (Souza et al., 2009). Therefore as the lateral wedge worn by the mature runners in the current study included a heel-based incline with little forefoot support, it could be suggested that position of the wedge was the possible reason for the non significant changes. It is therefore proposed that it is the location of the lateral wedge within the shoe that influences the effect on the kinematic movement patterns during gait. As the lateral wedge worn by the mature females was low thickness (6 mm) and based at the heel, the effect on peak rearfoot motion was limited.

The effect of the lateral wedge on the peak ankle dorsiflexion and peak forefoot abduction values similarly produced no significant changes. The change in peak forefoot abduction was negligible, with a mean difference of 0.1 degrees between the neutral and lateral wedge condition ($p>0.05$, effect size: 0.00006). The non significant change in forefoot abduction among the mature females was expected, due to the association between this variable of gait and pronation. The results from this study among mature females supports the coupling action between rearfoot eversion and forefoot abduction, with the lateral wedge producing non significant effects for both variables during running gait.

Currently, little research investigating the direct effect of lateral wedge technology on forefoot movement exists, however an association between forefoot varus and knee external adductor moment has been demonstrated (Teichtahl et al., 2006). According to Andrews and co-workers, participants that displayed increased forefoot varus had significantly lower knee external adductor moments during gait ($r = 0.44$, $p>0.05$) (Andrews, Noyes, Hewett, & Andriacchi, 1996). This theory was later supported by Hurwitz and colleagues, who showed forefoot varus to be significantly correlated with the knee external adductor moment during mid-late stance among osteoarthritis patients ($r = -0.45$, $p<0.001$), indicating that increases in knee moment reduce forefoot varus (Hurwitz, Ryals, Case, Block & Andriacchi, 2002). Therefore, as lateral wedge technology is designed to decrease the knee external adductor moment, it could be expected that an increase in forefoot abduction would be seen. However, this was not shown among the mature female runners in the current study, and it is suggested that the lack of significant results is due to the heel-focussed wedge in the shoe.

As no significant changes in rearfoot eversion and forefoot abduction were demonstrated by the mature females; the non-significant difference in peak ankle dorsiflexion with the lateral wedge was expected, due to the association of all three variables of gait. Ankle dorsiflexion is commonly linked with rearfoot eversion and forefoot abduction, as the first metatarsal joints lower in relation to the lateral border of the foot, forcing the latter into dorsiflexion (Bates et al., 1978). As such, it was expected that a lateral wedge would increase this angle, producing higher peaks of dorsiflexion among the mature females. However, the lack of significant difference is once again suggested to be due to the wedge acting at the heel or a consequence of the low wedge thickness (6mm).

5.4.2.3.2. Knee joint kinematics and moments.

The lateral wedge was expected to produce little effect on the maximum knee flexion angle during the stance phase of gait. Maximum flexion is achieved at mid stance, as the body weight is positioned over the foot (Kirtley, 2006). Among the mature females, no significant change was seen in the peak angle, time of occurrence or knee flexion angular velocity during the running trials performed in the lateral wedge. These results are similar to those presented in the literature. In one assessment of the effect of lateral wedges, Crenshaw and colleagues showed no significant change in knee flexion angle among a group of healthy individuals (Crenshaw, Pollo & Calton, 2000). This was later supported by Russell and co-workers who similarly showed a non significant change in knee flexion angle among healthy subjects, with mean peak angles of 15.5 and 15.8 degrees for the lateral wedge and control conditions respectively (Russell et al., 2003). Richard, Jones and Kim (2006) performed a later study with a brace and lateral wedge, and illustrated no change in knee flexion angle among participants. Although the dominance of studies involve assessments of walking gait, the consistency of results support those found among the mature females, and suggest little effect of lateral wedge technology on the knee flexion angle in the sagittal plane during gait.

Among mature female runners in the current study, the lateral wedge was predicted to significantly reduce peak knee abduction angle during the stance phase of gait. The mean

data illustrated a mean difference of 2.9 degrees between the neutral and the lateral footwear conditions, although this difference was not significant ($p>0.05$). However, with an effect size shown as 0.067, it is shown that 6.7 % of the variance in knee abduction was due to the lateral wedge; a moderate effect (Cohen, 1988). Based on the literature, and the results from the previously discussed kinematic variables, it is suggested that it could be the location of the heel wedge that limits the reduction in knee varus angle among this group. This is supported by Hinman and colleagues, who suggested that a full length orthotic with a wedge expanding the entire lateral border is more effective in affecting knee dynamics than a wedge at the heel (Hinman, Bowles, Payne, & Bennell., 2008). According to Inman (1976), this is due to the orientation of the subtalar joint axis; inclined from the lateral calcaneus to the first metatarsal head. As such, the full length lateral wedge supports the lesser metatarsal joints, and produces a lateral shift in the location of the centre of pressure throughout mid stance (Kakihana et al., 2004). This is further supported with the results of the investigation among the mature females, as the peak abduction angle occurred at a later time (0.22 seconds) than peak rearfoot eversion angle (0.096 seconds), suggesting the efficiency of controlling lower limb kinematics with the lateral heel is diminished shortly after heel strike.

The results from the knee internal rotation data among the mature females do not support the study hypotheses, with no significant increase in knee rotation with the lateral wedge. This result is similar to that produced by Nester and colleagues, who showed no change in peak knee internal rotation angle during gait with a lateral wedge orthotic (Nester el al., 2003). Due to the coupling mechanism between the rearfoot and the tibia, it is suggested that the non significant change in knee internal rotation is associated with the lack of change in rearfoot eversion shown with this footwear condition, and the height of the wedge could be the limiting factor in affecting the tibial and knee internal rotations during gait.

The kinetic knee data produced from the running trials in the lateral wedge however lend support to the second hypothesis. When compared to the neutral condition, the lateral wedge elicited a significant reduction in the peak knee external adductor moment, and moment arm length among the mature female runners. Therefore despite a consistent lack of kinematic change observed with the lateral wedge, this specific footwear condition did produce a significant reduction in the knee external adductor moment among mature

female runners (P>0.05). This is consistent with results presented in the literature, which have investigated the effect of lateral wedges on the knee external adductor moment and knee joint loading (Crenshaw et al., 2000). Although not a direct measure of knee joint loads, due to its strong association with medial knee loads this frontal plane knee moment often acts as a dependant variable when assessing effective strategies to reduce medial knee joint loading during gait (Hunt et al., 2007).

Lateral wedge technology is often used to modify the knee external adductor moment and decrease medial loads and osteoarthritis risk at the knee joint (Russell & Hamill, 2011). This is achieved through a shift in the centre of pressure laterally relative to the centre of the knee joint, which causes a decrease in the length of the moment arm of the ground reaction force about the knee joint centre (Yasuda and Sasaki, 1987). The results from the running trials performed by the mature females in the lateral wedge support this theory, as a significant reduction was seen in the peak moment arm calculated during the stance phase of gait (p<0.05). This therefore suggests that the wedge in the lateral heel and midfoot affected the moment arm produced at peak stance with enough magnitude to significantly affect the knee external adductor moment produced by the mature female runners.

When comparing these results with those presented in the literature, direct comparisons were compromised due to the lack of studies investigating lateral wedges during running. Despite this, one study performed by Nigg and colleagues demonstrated a 4.5 mm lateral wedge to significantly shift the centre of pressure laterally and reduce the moment arm length among subjects during running trials. It is however noted that this significant change in centre of pressure location was achieved only with a full length lateral wedge, and no significant changes were observed with the half length wedge (Nigg et al., 2003).

As suggested, the majority of research in this area has involved subjects performing walking trials, although many studies have included varying speeds, and as such the results from the faster speeds were primarily considered. Erhart and co-workers investigated the effect of two different heights of lateral wedge (4 degree and 8 degree) on a group of healthy individuals performing walking trials at slow, medium and fast speeds. Results from this study indicated a systematic increase in reduction of knee external adductor moment with both the height of the lateral wedge and the speed of walking; with walking trials in the 8 degree wedge producing a reduction of 19.66 Nm/kg

x m% at the fast speed compared with a reduction of 12.56 Nm/kg x m% at the normal speed (Erhart et al., 2008). Similarly, the effects of lateral wedge in walking trials at self-selected speeds have commonly shown decreases in knee external adductor moments (Crenshaw et al., 2000) although the results are not always consistent (Maly et al., 2002).

Additional studies support the results among the mature females, showing the effect of lateral wedge technology among healthy individuals. One study performed by Kakihana and co-workers showed a 6 degree lateral wedge to decrease the knee external adductor moment by an average 24.44 %, and a later investigation showed a 5 degree lateral wedge to decrease the peak knee external adductor moment by an average of 6.76 % (Kakihana et al., 2004; Crenshaw et al., 2000). Conversely, Maly and colleagues indicated a 5 degree lateral wedge to produce no change in knee external adductor moment during walking. However it is noted that this investigation was performed using a heel wedge, and subjects were all diagnosed with medial compartment knee osteoarthritis (Maly et al., 2002). In the same way, Nester and co-workers showed minimal effect of lateral wedges on the knee kinetics and kinematics among healthy subjects during walking trials, although similar to the previously discussed study (Maly et al., 2002; Nigg et al., 2003), the lateral wedges were not full length (Nester et al., 2003).

Although the results among the mature females indicated a positive effect of lateral wedges, the inconsistency of significant changes seen in knee kinetics with a lateral wedge is discussed by Shelbourne and colleagues. Here it is suggested that a lateral wedge will cause a lateral shift in the centre of pressure. However, even a substantial displacement of centre of pressure of 5 mm in the lateral direction, will only produce a 3 Nm (0.5 % BW * height) decrease in knee external adductor moment, suggesting a possible reason for often non significant changes in peak moment (Shelbourne et al., 2008). Therefore, the majority of research, combined with the results among the mature females lend support to the theory that a lateral wedge will reduce the knee external adductor moment during gait. It is however clear that an increase in research on the effect of these wedges during running gait is required to substantiate these theories.

5.4.2.4. Orthotic footwear condition; lateral wedge and medial support.

Among the mature females, the running trials performed in the orthotic condition elicited the biggest change in both kinematics and kinetics of the lower limb. It was hypothesised that the medial support in the orthotic would significantly reduce rearfoot eversion and knee internal rotation, while the full length lateral wedge would produce reductions in knee abduction and peak knee external adductor moments.

5.4.2.4.1. Orthotic induced kinematic changes at the ankle/subtalar joint.

At the ankle joint, the orthotic footwear condition produced minimal changes in peak dorsiflexion angle and peak foot abduction during running, with slight reductions demonstrated in both angles ($p>0.05$ effect sizes: 0.084, 0.0021). The peak rearfoot eversion angle was however significantly reduced among the group when compared to the neutral condition ($p<0.05$), although no change was seen in the peak eversion occurrence time or velocity ($p>0.05$). These results initially tend to contest the theory that foot abduction and dorsiflexion are linked with rearfoot eversion. Although the slight reductions in both kinematic variables do support the theory that rearfoot eversion is the primary variable of pronation (Bates et al., 1978). Additionally, although designed to reduce the negative effects of pronation, the medial support in the orthotic condition was primarily hypothesised to reduce rearfoot motion.

The significant change in rearfoot eversion angle among mature females supports the theories in the literature, suggesting increased medial support in the heel and the arch reduce the maximum rearfoot eversion angle during the stance phase of gait (Kilmartin et al., 1994). In a review of literature, Kilmartin and colleagues suggest that the majority of research reports a similar finding; that medial support biomechanical orthoses reduce rearfoot pronation during gait (Kilmartin et al., 1994).This supports Novick and Kelley (1990), who examined the effect of orthotics with postings in the rear and forefoot, and showed a significant decrease in peak calcaneal eversion with respect to the lower leg. Similarly, Eslami and co-workers showed a medial support foot orthotic to significantly

reduce peak rearfoot eversion angle during running, with no change in occurrence time, a result similar to that produced by the mature females (Eslami et al., 2009). MacLean and colleagues also supported this theory, showing an orthotic footwear condition to significantly reduce maximum rearfoot eversion angle among healthy runners (MacLean et al., 2006). The results from the mature females therefore lend support to these previous studies. Despite this, there is still uncertainty based around the role and efficiency of orthotics and previous research has also shown little effect of medial posting on rearfoot motion during running and walking. Stacoff et al (2000) reported small, non significant and inconsistent effects of orthotics on the rearfoot mechanics of healthy individuals, and Stackhouse et al (2004) later supported this, showing foot orthotic devices to produce no significant effect on rearfoot motion among healthy subjects performing running trials. However it was noted that although the overall group response was not significant, some subjects did show a reduction in peak rearfoot angle, supporting the theory that individual responses to orthotics can hinder significance of group studies (Stackhouse et al., 2004).

Due to the association between excessive rearfoot eversion and injuries, the effect of specific footwear and orthotic interventions on rearfoot motion during running has received a high level of interest, with research consistently producing equivocal results. The results from this investigation among the mature females support the use of medial wedging to reduce rearfoot eversion among mature female runners. Additionally, the inclusion of a lateral support within the shoe elicited no negative effects on the subtalar joint movement, a notion supported by Nakajima and colleagues who showed a combined orthotic similar to the one worn by the mature females to produce a reduction in rearfoot eversion (Nakajima et al., 2009). The lack of negative effect on the rearfoot eversion was further supported by Teichtahl and colleagues, who suggested that at early stance the ground reaction force acts through the heel and therefore the position of the forefoot has little effect on the knee kinetics (Teichtahl et al., 2006). This is an important result as the magnitude of peak ankle eversion has implications for soft tissue injury, particularly at the knee (Bates et al., 1979; Tiberio, 1987). It appears that the study orthotic device could therefore act as a successful strategy for reducing key biomechanical variables associated with knee injury.

The lack of significant change in rearfoot eversion velocity is not surprising due to the mixed responses presented in the literature. In an early study, Smith and colleagues

showed foot orthotics to produce a decrease in rearfoot eversion velocity, however Brown and colleagues later showed no effect of orthotics on eversion velocity while walking, and Stackhouse later showed no change in calcaneal velocity with orthotic intervention when running (Smith Clarke, Hamill and Santopietro, 1986; Brown, Donatelli, Catlin, & Wooden, 1995; Stackhouse et al., 2004). It is therefore suggested that although the peak angle of rearfoot eversion may change with orthotics, the velocity and occurrence time may not be affected.

When compared with the neutral condition, the orthotic produced a slight decrease in ankle dorsiflexion (effect size: 0.084, medium). As a component of pronation, a reduction in ankle dorsiflexion was expected, as the orthotic was intended to reduce pronation values. The lack of significant change in dorsiflexion angle is suggested a possible consequence of data collected from only 20 females, as 80% of the participants showed a reduction in dorsiflexion value with the orthotic. This could therefore be a result of individual variability in response to the orthotics, as four out of the twenty participants showed an increase in peak ankle dorsiflexion with the orthotic condition. The increase in peak ankle among four mature females suggests an independence of ankle dorsiflexion movement from rearfoot eversion, and could be due to the orthotic elevating the medial arch, which increases the mediolateral position of the subtalar axis, and produces an increase in dorsiflexion about this axis (Stackhouse et al., 2004). However this response may be expected with only medial support, and the lateral support in the orthotic may have counteracted this movement.

The overall lack of significance in ankle dorsiflexion movement with the orthotic is suggested a consequence of the combined medial-lateral support, reducing the overall movement of the ankle in the sagittal plane, and variability in individual response. This is similar to the results produced for the forefoot abduction variable, where no significant changes occurred between the neutral and the orthotic condition among the mature female runners. It has previously been suggested that the inclusion of a lateral wedge reduces the forefoot abduction angle during gait (Nester et al, 2003); however this is reversed by the inclusion of a medial wedge within the orthotic (Nakajima et al., 2009). Therefore the arch support included in the orthotic worn by the mature females could have counteracted any reduction in foot abduction produced by the lateral wedge technology. According to Hurwitz and colleagues, an increase in toe out gait (foot abduction) produces a significant

reduction in the knee external adductor moment, and subsequent medial knee joint loading. This therefore initially suggests a possible limitation to the lateral wedge intervention designed to reduce this moment. However, both the results among the mature females, and those presented by Nakajima and colleagues suggest that the inclusion of medial support within the orthotic device counteracts this possible drawback, with no significant reduction in foot abduction shown (Nakajima et al., 2009). This is further supported with the kinetic results that will be discussed in a later section, displaying no increase in knee external adductor moment with the orthotic intervention.

At the ankle joint, the orthotic intervention significantly reduced rearfoot eversion, but produced no change in ankle dorsiflexion or foot abduction. This tends to fall in line with previous research, and the group significance demonstrated suggests a positive role of this form of orthotic among mature females.

5.4.2.4.2. Orthotic induced changes in kinematics at the knee joint.

At the knee joint, the orthotic condition was hypothesised to significantly reduce knee internal rotation and knee abduction angle. Overall, little change was demonstrated in the peak knee flexion or abduction angle, although the orthotic intervention significantly delayed the occurrence time of the peak knee abduction angle ($p<0.05$). Additionally, the orthotic significantly reduced the magnitude of peak knee internal rotation angle and angular velocity during the stance phase of running gait ($p<0.01$).

According to Nawoczenski and colleagues, knee flexion is linked with subtalar joint pronation and tibial and knee internal rotation in the weight-bearing position (Nawoczenski et al., 1995). It is possible that the increase in support on the lateral sole counteracted this effect, reducing the significance of knee flexion reduction. The lack of significant change in peak knee flexion angle among the mature female runners was similar to the early work of Cavanagh and Edington, who showed both a 10 degree lateral and medial wedge to produce no effect on the magnitude or time of peak knee flexion during the stance phase of gait, and Bates and colleagues who showed no effect of orthoses on knee flexion during treadmill running (Cavanagh & Edington, 1989; Bates et

al., 1979). This was further supported more recently by Nester and colleagues who similarly showed no change in knee flexion angle with orthotic intervention (Nester et al., 2003). It is possible that as the wedging in the orthotic worn by the mature females was based on both the medial and lateral sides of the sole, the effect on kinematics in the sagittal plane of motion is minimal. This is further supported by the lack of significant change in the peak ankle dorsiflexion angle.

Although no significant change in knee abduction angle was elicited with the orthotic condition, the peak angle was reduced in 19 of the 20 participants, with an average reduction shown as 3.3 degrees. Additionally, 6.5 % of the change in knee abduction was due to the orthotic condition (effect size: 0.065). It was expected that the lateral wedge would decrease the peak abduction angle among the mature females; however the inclusion of the medial support may have limited the magnitude of this reduction. Although the difference in magnitude of peak knee abduction angle was not significant, it is suggested that the reduction in knee abduction angle is associated with the reduction in moment arm length at the knee joint. A displacement of the knee joint centre towards the midline of the body reduces the moment arm between the centre of the knee joint and the line of the ground reaction force acting vertically (Hunt et al., 2006). Therefore, although not significant, the orthotic condition is suggested to have a greater effect on the kinematics at the knee joint through the decrease in knee abduction angle during the stance phase of gait. It is possible that the significance of this reduction in knee abduction angle could be influenced by the number of participants (20) within this current study, and an increase in participants may have substantiated this result. This is supported by statistical power analysis, and based on the 0.9 degree change in knee abduction angle with the motion control shoe in Study 2, a sample of 20 females would only elicit a power of 17%. However, this was based on results with a motion control shoe, with no specific intervention designed to alter knee abduction. Based on the means from the current study, the effect of the orthotic intervention on the knee abduction would produce a power of 54 % with a sample size of 20. Therefore, although higher than the power from the previous study, this could be considered a limiting factor.

Despite no significant change in peak angle, the orthotic, combining lateral wedge and medial support did however produce a significant delay in the peak knee abduction occurrence time among the mature females, showing a maximum angle to occur 0.02

seconds later than in the neutral condition. This, combined with a reduction in angular velocity, suggested that the lateral support in the shoe delayed the movement of the knee to peak abduction, possibly reducing the speed with which load is dispersed to the medial compartment. This therefore could also be related to the reduction in knee external adductor moment that occurred during late stance.

The main finding among the kinematic data at the knee joint is the significant reduction in peak knee internal rotation angle during running. As illustrated, the medial wedge within the orthotic condition significantly reduced the rearfoot eversion angle among the mature females. Due to the coupling between tibial and knee internal rotation with rearfoot eversion, the decrease in rotation at the knee joint was expected. This falls in line with previous research showing the positive effect of orthotics on the knee internal rotation angle during gait (Nawoczanski et al., 1995; Stackhouse et al., 2004). However the main interest in the results among the mature females is the lack of negative effect of the lateral wedge within the orthotic device. It was possible that the lateral wedge would counteract the medial support, enabling the tibia and knee to internally rotate, as was demonstrated in the neutral condition. However the addition of the lateral support still produced a significant reduction in peak knee internal rotation angle during running. It is suggested that as the tibial and knee internal rotation is produced by the motion of the rearfoot, the medial wedge based at the heel and medial arch limit this motion. As the majority of lateral support is based in the mid and forefoot, this support has little effect on the magnitude of knee internal rotation achieved during running. The significant reduction of 0.43 radians per second in knee internal rotation velocity further supports these theories, showing the medial support to reduce the speed with which the knee internally rotates during stance.

5.4.2.4.3. Effect of orthotic intervention on the knee external adductor moment.

Among the mature females, the running trials performed in the orthotic condition elicited the greatest reduction in knee external adductor moment, with an average decrease of 0.23 Nm/kg from the neutral condition. These results support the study hypothesis for this

variable with the reduction showing statistical significance ($p<0.01$). The peak knee external adductor moment was measured from the second peak (Figure 5.22), and therefore occurred during mid to late stance. To this end, it was expected that the lateral wedge based along the length of the lateral sole would restrict movement of the knee joint towards a varus position at mid to late stance, thereby reducing the moment arm at the knee joint. These results were further supported by the significant reduction in moment arm length, measured from the midpoint of the knee joint to the line of frontal ground reaction force acting through the joint, ($p<0.05$). The moment arm was also significantly correlated with the peak knee external adductor moment in the neutral, orthotic and lateral footwear conditions ($p<0.05$).

These results fall in line with many studies in the literature supporting the use of lateral wedge technology for reduction in knee external adductor moments and consequential medial knee joint loading (Crenshaw et al., 2000; Russell et al., 2011). However, a main point of interest when discussing results produced in the orthotic intervention is the lack of negative effect of the medial wedge within the orthotic device. During the stance phase of gait, the heel strikes in an inverted position, loading the lateral side of the calcaneus (Whittle, 2003). As gait progresses to early midstance, the calcaneus moves quickly into eversion; a movement closely coupled with internal rotation of the tibia and knee joint (Richards, 2008). At this point, the knee often moves into an adducted position. Among the mature females, as gait progresses into mid to late stance, the knee joint moves into abduction, creating a moment arm between the centre of the knee joint, and the medio-lateral coordinate of the centre of pressure, and a knee external adductor moment acts at the joint (Figure 5.1). This progression of movement therefore lends explanation to the positive results displayed by the medial and laterally wedged orthotic worn by this group. As the medial wedge is based at the heel and arch, this component controlled for early stage gait movements including rearfoot eversion and knee internal rotation. However as the lateral component is based along the mid and forefoot, the abduction movement and knee external adductor moment is limited.

The significant correlation between moment arm length and knee external adductor moment in the neutral, orthotic and lateral footwear conditions, lends support to the theory that altering alignment and moment arm length is an effective way to reduce the knee external adductor moment during running (Hunt et al., 2006). These results are

supported by a recent study within the literature. Nakajima and colleagues investigated the effect of a lateral wedge with medial arch support among a group of healthy individuals. In the results it was shown that although no change in the peak knee external adductor moment was shown with the inclusion of medial arch support, the average moment throughout stance was significantly reduced, with the greatest reduction shown at late stance (Nakajima et al., 2009). In contrast, Abdallah and co-workers investigated the effect of laterally wedged orthotics with medial support on the walking gait of patients diagnosed with medial compartment knee osteoarthritis. Results indicated that no significant difference was seen in the knee external adductor moment, (Abdallah et al., 2009). However, although no change was demonstrated in the knee external adductor moment with the lateral wedge, the addition of a medial wedge did not increase the knee adduction moment. Furthermore, on evaluation of the methodology, it is noted that these results were displayed among a group of diagnosed patients, and according to Kakihana and colleagues, osteoarthritis patients demonstrate significantly smaller changes in kinetics and kinematics with orthotics than changes seen among healthy participants (Kakihana et al, 2007).

5.5. Conclusion: Positive effect of combined medial and lateral support.

Although lateral wedge technology has been advocated to reduce knee external adductor moments and medial knee joint loading, a lack of support under the medial arch while placing the foot in an everted position increases the tension and peak pressures on the medial aspect of the foot (Butler et al., 2007; Tsung et al., 2004). Similarly, many subjects have complained of discomfort when walking in shoes with lateral wedges (Abdallah et al., 2011). As such, the addition of medial support at the rearfoot and medial arch is suggested to reduce these negative effects (Nakajima et al., 2009).

The results from the running trials performed in the orthotic footwear condition in this study support two main theories presented in the literature. Firstly, that the length of the lateral support influences the effect on knee kinetics, and secondly, that the addition of a medial arch support improves the overall effect of a laterally wedged orthotic (Hinman et al., 2008; Nakajima et al., 2009). Results from both the individual medial and lateral wedges showed little significance on the kinematics among mature female runners. However, both the full length lateral wedged orthotic intervention, combined with the addition of a medial wedge at the heel and arch produced significant reductions in gait variables commonly associated with debilitating conditions.

Both rearfoot eversion and knee internal rotation have been related to the development of lower limb and knee injuries and overuse conditions, due to an increase in strain on the ligaments and an uneven force distribution at the knee joint (Cheung et al., 2006). Similarly, the knee external adductor moment is a variable of gait that has been consistently related to the development of medial compartment knee osteoarthritis. Within this study, the orthotic device produced significant reductions in both peak rearfoot eversion and knee internal rotation angle among a group of mature female runners, with no negative influence of a lateral wedge seen among this group. Furthermore, the significant reduction in the knee external adductor moment produced with the orthotic device supports the theory of lateral wedge technology with combined medial support in potential management of overuse injuries and medial compartment osteoarthritis at the knee joint.

Although the orthotic condition did not produce significant changes among all investigated variables, including knee abduction angle, the lack of consistency of results could be due to the variability in response to orthotic devices and the relatively small sample sizes. This is supported by the moderate effect sizes shown. To see the same response in knee abduction among 19 of the 20 mature females is potentially unusual and although many results were not significant it is possible that a larger sample size would increase likelihood of significant results (Nigg et al., 1999).

The results from this study among mature female runners therefore suggests that an orthotic intervention combining both a lateral wedge with medial rearfoot and arch support could significantly reduce the risk of lower extremity injuries and medial compartment knee osteoarthritis among this group.

Chapter 6

"Biomechanical Indications of Medial Knee Osteoarthritis and Evidence of Bone Changes on Magnetic Resonance Imaging."

6.1. Introduction and Review of Relevant Literature.

6.1.1. Longitudinal changes to biomechanics for six consistent mature female participants.

Within this thesis, three previous biomechanical investigations have included a subset of six constant participants. As such, a review of kinematic results produced by these mature females provides an indication into longitudinal changes over the course of 30 months. Table 6.1 indicates the changes in rearfoot eversion angle for the 6 participants over this duration, and Figure 6.1 indicates the similar changes in the knee external adductor moment for each participant.

Table 6.1. Mean rearfoot eversion and knee internal rotation angle produced by the 6 participants within each biomechanical investigation.

PARTICIPANT	REARFOOT EVERSION (deg)			KNEE INTERNAL ROTATION (deg)		
	Study 1	Study 2	Study 3	Study 1	Study 2	Study 3
1	11.2 ± 3.49	14.1 ± 3.12	14.5 ± 3.9	14.0 ± 2.4	16.0 ± 3.2	17.4 ± 1.23
2	12.5 ± 1.6	15.9 ± 3.62	18.5 ± 5.7	16.0 ± 2.5	10.8 ± 3.28	9.5 ± 1.1
3	10.5 ± 3.2	12.3 ± 1.2	13.1 ± 6.42	12.6 ± 3.5	12.6 ± 2.75	12.5 ± 1.91
4	12.9 ± 4.1	17.1 ± 2.79	17.2 ± 2.82	14.5 ± 3.49	14.0 ± 3.42	14.0 ± 2.4
5	13.4 ± 2.38	18.1 ± 8.17	19.6 ± 3.22	13.8 ± 3.2	14.8 ± 1.53	12.0 ± 3.99
6	14.0 ± 6.1	21.2 ± 13.5	19.9 ± 2.67	14.5 ± 2.8	11.6 ± 3.90	14.3 ± 2.3

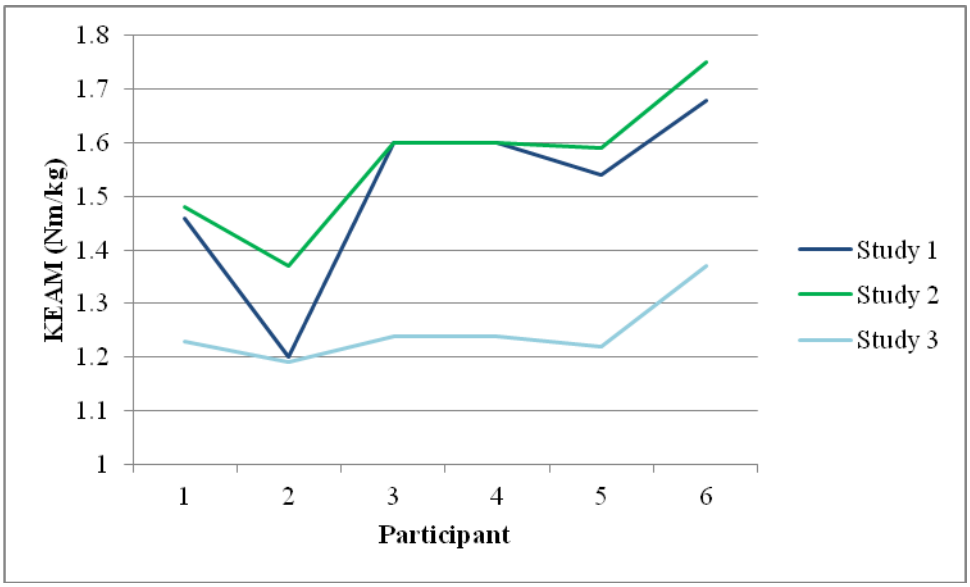


Figure 6.1. Knee external adductor moment (Nm/kg) produced by six participants across the three biomechanical investigations.

Although there is a tendency for both the rearfoot eversion angle and the knee internal rotation angle to increase across this period, there is a notable decrease in the knee external adductor moment among each participant in the final biomechanical investigation. As the footwear was changed for each study, it is possible that the shoes worn in the final study incorporated certain design features that decreased the varus position of the knee joint, and decreased the knee external adductor moment acting at the joint. It is however noted that the inter-individual patterns remained similar, suggesting the knee external adductor moment was reduced similarly among each participant.

Due to the general increase in rearfoot eversion and knee internal rotation angle across the three studies, and the increase in the knee external adductor moment across studies one and two, it is suggested that gait changes occur with respect to increasing age, and the vulnerability of the knee joint to degenerative conditions has increased. Despite a lack of increase in the knee external adductor moment in the third biomechanical investigation, the moments produced remain high when compared to literature examples, and therefore suggest the risk of knee joint osteoarthritis remains high among this specific group.

6.1.2. Osteoarthritis; a debilitating condition of the knee joint.

Osteoarthritis is a major cause of pain and mobility-related disability among an ever increasing number in the population (Felson et al., 2003). This condition is commonly considered a consequence of the natural wear and tear of synovial joints over time, is most frequently present in the knee joint, and is often preceded by an injury (Tiderius et al., 2003). However, advances in laboratory and clinical research has recognised osteoarthritis as a complex mechanical and biological disorder of the joint that in time leads to loss of function and pain (Hascall & Kuettner, 2002).

6.1.2.1. The knee joint and articular (hyaline) cartilage.

The knee joint is the largest and most complex joint in the human body, comprising three joints within a single synovial cavity (Totora & Grabowski, 2003). The details of the anatomy and mechanics of the knee joint are provided in Chapter 2; however a brief summary is provided here to, due to the nature of this study. Simply, the knee encompasses the distal end of the femur, the proximal end of the tibia, and the patella. Laterally, the knee forms the tibiofemoral joint, comprising the lateral condyle of the femur, lateral meniscus and lateral condyle of the tibia. Conversely, on the medial side, the second tibiofemoral joint comprises the medial condyle of the femur, the medial meniscus and the medial condyle of the tibia. Both tibiofemoral joints are hinge joints. Finally, the intermediate patellofemoral joint is formed between the patella, and the patellar surface of the femur, forming a planar joint (Totora & Grabowski, 2003).

At the knee joint, articular (hyaline) cartilage lies over the articulating surface of the distal femur and proximal tibia, and the posterior side of the patella. This is the most common form of cartilage, composed of cellular chondrocytes and an extracellular matrix made up of hyaluronic acid, proteoglycans, type II collagen, and water (Majumdar, 2010). In addition to this articular cartilage, menisci are also present at the knee joint, consisting of fibrocartilage, a dense, fibrous and more resistant form of cartilage, arranged in concentric plates in the medial and lateral knee joint (Totora & Grabowski, 2003).

6.1.2.2. Knee joint osteoarthritis development.

The pathogenesis of osteoarthritis is manifested by a combination of morphologic, biochemical, molecular, and biomechanical changes of both the cells and matrix of the structures and articular cartilage at the knee joint (Majumdar, 2010). This leads to softening, and fibrillation of the articular cartilage, eburnation of the subchondral bone, and the development of osteophytes and subchondral cysts (Moskowitz, 2007). During the development of osteoarthritis of the knee joint, the loss of hyaline cartilage is the central pathologic event (Felson et al., 2003). Articular cartilage is gradually lost due to a developed imbalance between biosynthesis and degradation of matrix constitutes such as type II collagen and glycosaminoglycans (Tiderius et al., 2003). During the early stages of osteoarthritis development, when compared with healthy cartilage, the degenerating hyaline cartilage contains increased water and decreased proteoglycan concentrations. This is coupled with weakened collagen network, due to the decreased synthesis of Type II collagen fibres (Majumdar, 2010). These changes have been attributed to increased presence of specific molecular messengers such as Interleukin-1 within the articular cartilage, which releases catabolic enzymes and acts to suppress the synthesis of type II collagen. It has also been shown that the chondrocytes of osteoarthritic knee joints are increasingly susceptible to Interleukin-1 than are those found in healthy joints (Majumdar, 2010).

Following the breakdown of collagen within the articular cartilage, the matrix undergoes a process of fibrillation as the superficial layers are broken down. Eventually, the cartilage is worn down through the full depth; exposing subchondral bone plates which become the new articulating surface of the knee joint (Majumdar, 2010). As such, although cartilage loss is a known pathologic function of osteoarthritis development, abnormal bone formation is another associated element (Felson et al., 2003). As normal everyday movement occurs at the knee joint, the stages of osteoarthritis progress, and the exposed bone undergoes repeated frictional forces, and becomes smoothed and burnished (eburnation). This then leads to the formation of sub-chondral cysts and osteophytes, established in the areas of exposed bone.

Bone edema (bruising) is an additional feature, caused by changes in articular contact points at the sites of biomechanically failing cartilage (Moskowitz, 2007). Bone edema

has been repeatedly illustrated as a feature of osteoarthritis, and is associated with increased pain and risk of further cartilage degradation (Felson et al., 2001). Histologically, this feature of osteoarthritis reflects pathologic evidence of increased water, blood or other fluid within the bone, and is often unrecognisable in radiographic diagnosis of the condition. A diagram of the osteoarthritic knee is illustrated in Figure 6.2.

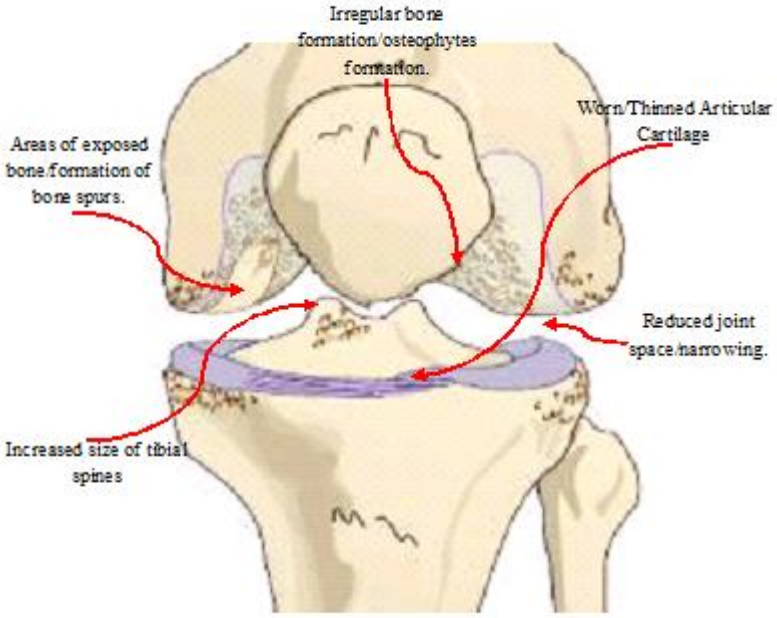


Figure 6.2. Adapted simplistic illustration of the right knee displaying osteoarthritic changes. Source: sportsinjuryclinic.net.

6.1.3. Epidemiology of osteoarthritis.

Osteoarthritis is a degenerative condition that affects an ever expanding number among the population. As the most common skeletal disorder in adults, the risk of osteoarthritis increases with age, and is almost twice as common in women as it is in men. These gender differences are often attributed to the hormonal influence of oestrogen on the structure of the articular cartilage, a gender variation in injury risk, and differences in biomechanics between males and females, including increased Q angle among females, and a lower muscle strength to body mass ratio (Moskowitz et al., 2007). Several studies have supported these theories, with specific statistics showing osteoarthritis of the knee joint to be more prevalent in women than their male counterparts (Davis, Ettinger., & Neuhaus, 1989).

The risk of developing osteoarthritis at the knee joint is also increased in those who have suffered earlier trauma to the joint (Zuluaga, Briggs, Carlisle, McDonald, McMeeken, Nickson, Oddy & Wilson, 1995). Ligamentous and meniscal damage decreases stability of the knee joint, and leads to excessive and uneven loading in specific areas. Therefore, with higher force being attenuated at the knee joint in activities such as running, the uneven distribution of load heightens the risk of structural degeneration and osteoarthritis development (Zuluaga et al., 1995).

6.1.4. Biomechanical alignment associated with knee osteoarthritis.

As well as age and gender, lower limb malalignment has also been reported as a potential risk factor for the development and structural progression of knee osteoarthritis (Felson et al., 2003). This condition is a result of local mechanical factors acting within the context of systemic susceptibility at the knee (Sharma et al., 2001). At this joint, site-specific factors, including alignment, govern how load is distributed across the structure, and a disproportionate medial transmission of load occurs from a stance phase adductor moment. As discussed in the previous chapter, the knee external adductor moment, and increased moment arm length, is associated with a varus position at the knees (Figure

5.1). This moment has been shown to reflect the intrinsic compressive load acting on the medial compartment during gait (Sharma et al., 2001). One longitudinal study illustrated an increased risk of medial compartment osteoarthritis progression among patients with knee varus alignment; baseline knee varus was associated with a four-fold increase in the risk of medial compartment progression (Sharma et al., 2001). These findings are consistent with previous biomechanical studies that have shown increased knee varus alignment to increase medial compartment load and risk of progression (Hsu et al., 1990; Miyazaki et al., 2002).

It is however often questioned whether increased knee external adductor moment and varus alignment precedes osteoarthritis development, or is simply a consequence of the condition acting at the medial knee (Brouwer et al., 2007). Osteoarthritis present at the knee joint can cause malalignment due to articular cartilage loss and changes in bone formation, resulting in an uneven articulating surface (Sharma et al., 2001). Brouwer and colleagues investigated this theory and illustrated lower limb varus malalignment, measured as the femorotibial angle using radiographic imaging, to predate knee osteoarthritis development (Brouwer et al., 2007). As such, it is assumed those mature females that present with high knee external adductor moments in the current study are at an increased risk of developing osteoarthritis in the medial compartment of the knee joint.

6.1.5. Magnetic Resonance Imaging as a method to identify early stage osteoarthritis.

There is known to be a poor correlation between pathological severity of osteoarthritis and patient symptoms, and as such clinicians have constantly relied on medical tools for diagnostic purposes (Ehrman, Gordon, Visich, & Keteyian., 2003). Early stages of osteoarthritis have predominantly been identified using radiographic images, which depicts joint-space narrowing and osteophytes formation (Ding et al., 2007). However radiographic images have been shown to have a range of limitations in osteoarthritis identification, including limited sensitivity to change, and poor prediction of cartilage loss (Cicuttini et al., 2004). Furthermore, it has been illustrated that reduction in the volume of cartilage, alongside the development of smaller cartilage defects, has the ability to occur

within the joint before radiographic scans can identify abnormalities (Ding et al., 2007). As such, the use of MRI scans has been advocated to enhance identification of early signs of knee osteoarthritis, within both symptomatic and asymptomatic patients (Pessis et al., 2003; Cicuttini et al., 2004).

MRI can offer a distinct advantage over knee radiography, since cartilage can be assessed directly (Amin et al., 2005). Although radiography is non-invasive and inexpensive, it remains an indirect measure of cartilage loss through osteoarthritis. However in addition to cartilage changes, osseous modifications are ideally depicted by MRI, with visualisation of both cortical and trabecular bone. Furthermore, due to the tomographic nature of MRI scanning, this method is more advances at delineating structures such as subchondral osteophytes that are often obscured by overlying structures on conventional radiographs (Moskowitz, 2005). Standard MRI techniques including fat-saturated, T1-weighted and T2-weighted spin sequences have recently been utilised to directly assess structural changes in the knee joint, including cartilage volume, defects, subchondral bone changes and meniscal lesions, all which can be identified during the early stages of osteoarthritis development (Ding et al., 2007). Both T1 and T2 weighted scans are similar basic scans, which differentiate fat from water to highlight differences in composition of cartilage and muscle. In the T2 weighted scan, the fat shows up darker, and water lighter, highlighting suitability to imaging bone edema. Furthermore, MRI can provide 3-dimensional imaging of the knee, in contrast to the 2-dimensional radiographic images. As such, cartilage morphology can be assessed in multiple compartments of the knee joint. Commonly, the tibiofemoral joint is divided into five compartments of cartilage when assessing for osteoarthritis development using MRI. These include the central and posterior femur, and the anterior, posterior and central tibia (Amin et al., 2005).

One study investigating the superiority of MRI scanning above the use of radiographic images, examined the relationship between joint narrowing on radiographic images and cartilage loss on MRI (Amin et al., 2005). Coronal, axial and sagittal images of the knee were obtained from 224 men and women (mean age 66 years). Results showed that 46% had cartilage loss visible on MRI. Of this value, 42% of patients showed cartilage loss visible on MRI, where no radiographic progression was present. As such, it was concluded that radiography is not a sensitive measure of knee osteoarthritis, and if used unaccompanied by MRI scans, will miss a substantial proportion of knees with cartilage

loss (Amin et al., 2005). Furthermore, although changes in joint space are often detectable through radiographic images, it has been shown that these changes reflect cartilage loss in the central regions of the weight-bearing surface of the knee joint rather than any other areas (Amin et al., 2005). As such, MRI scans in the coronal plane could be used to facilitate identification of cartilage loss occurring at other areas of the tibiofemoral joint, particularly the posterior femur compartment.

According to Bartko (1995), vital characteristics of the ideal method to assess osteoarthritis of the knee joint include simplicity, validity, reliability, and sensitivity to change. The final characteristic is the most difficult to achieve, as sensitivity to change in cartilage breakdown over the course of one year is commonly insignificant through the use of radiography and arthroscopic parameters (Pessis et al., 2003). Arthroscopy can be used to analyse pre-radiographic cartilage changes, however as an invasive measure it results in post operative functional limitations and further complications (Tiderius et al., 2003). In an assessment of knee osteoarthritis progression, Pessis et al (2003) showed no statistically significant changes in cartilage degradation through the use of radiography and arthroscopy. However, in the same study, MRI scans in the coronal and sagittal planes illustrated significant worsening of the chondropathy, demonstrating the ability of MRI to identify characteristics of osteoarthritis that are unrecognisable through the use of radiographic x-rays (Pessis et al., 2003). Additionally, the presence of bone oedema that is recognisable through the use of MRI scans could determine whether the chondral disorder is likely to progress over the following year.

Bone marrow oedema is recognisable through coronal scans showing increased signal on the marrow on fat-suppressed T2-weighted images. These bone marrow lesions are defined as areas of increased signal, present in either the medial or lateral compartment of the femur or the tibia, and graded based on size. Previously it has been suggested that only lesions graded 2 and above are associated with pain. As such, within this study it would be pertinent to look for smaller lesions (graded below 2), as all participants are anticipated to be pain free (Felson et al., 2003). A study by Felson et al (2003) confirmed that bone lesions identified through MRI images significantly predicted local structural deterioration. When medial compartment lesions were identified, risk for medial compartment progression of osteoarthritis was increased more than six fold (Felson et al., 2003). Furthermore, the presence of lesions in the knee joint was commonly associated

with biomechanical malalignment, both statically and dynamically; with increased lesions in the medial compartments significantly associated with varus alignment of the lower limb. This suggests the high influence of frontal plane malalignment in the structural progression of knee osteoarthritis (Felson et al., 2003). Furthermore, this supports the suggestion that radiographic outcome varies with the specific degree of knee flexion (Tiderius et al., 2003). Therefore the above highlights the importance of using MRI images to identify bone marrow oedema seen in bone lesions (Felson, et al., 2003). These lesions have been shown to be present without the onset of pain, and have consistently been associated with the progression of osteoarthritis.

The weight of the above research therefore highlights the importance of MRI scans in the detection of early stage osteoarthritis of the knee joint. Although radiographic images have been used previously, it has been highlighted that changes in the morphology of the cartilage can occur prior to radiographic identification. Furthermore, MRI scans have been shown to identify progressive changes in the cartilage degradation over the course of one year, which appears untraceable through the use of radiography. MRI offers a safe and accurate method to assess early stage osteoarthritis non-invasively (Tideruis et al., 2003).

6.1.6. Aims and hypotheses.

The main aim of this investigation was to support the original theory that mature female runners are at a higher risk for the development of debilitating conditions, specifically knee joint osteoarthritis. This occurred with an assessment of the presence of early stage, medial compartment knee osteoarthritis among a group of asymptomatic mature female runners using magnetic resonance imaging. Two hypotheses were addressed:

1. Mature females would demonstrate early stages of knee osteoarthritis identifiable through the use of MRI.
2. A significant positive correlation will exist between the knee external adductor moment (as measured in the previous study) and the severity of early stage osteoarthritis determined from the MRI scans.

6.2. Methods.

6.2.1. Participant selection.

This investigation was provided ethical approval from the University of Exeter ethics committee. 10 female participants were recruited on a volunteer basis for participation in an investigation of early stage osteoarthritis of the knee joint. The females had a mean age of 51.8 years, with a minimum age of entry of 40 years, based on the methodology of the previous investigations. All participants had taken part in the biomechanical assessment involved in Study 3, which had occurred two weeks earlier. Therefore, the time delay between the data collected from the motion analysis investigation and the MRI assessment was minimal. As described previously, all females were classified at the same standard in running, by weekly club attendance and 10 km run time. All participants were asymptomatic and had never been diagnosed with knee joint osteoarthritis. All females provided informed consent, and completed a patient safety checklist, which assessed suitability for participation. Prior to this, each female was provided an information sheet, detailing the intentions of the study, the technology involved in each scan, and the low level risks apparent. Each female was requested to attend the scanner following a rest day, to limit the presence of knee joint bruising (bone oedema) possible following a run.

6.2.2. Magnetic Resonance Imaging.

All scans were performed with a Philips 1.5 Whole Body Imager (Philips Gyroscan Clinical Intera, Koninklijke Philips Electronics N.V), located in the Peninsular Medical Building at the University of Exeter. Each participant volunteered for a single session, during which scans were performed on the right knee of each mature female. All scans were performed with a Quadrature Knee Coil to hold the knee in place while in the scanner, and ensure uniformity between patients. The type of scan selected was a T-1 weighted scan with fat suppression, which differentiates fat from water. Due to the short repetition time, this standard basic scan enables prompt collection of high resolution 3-dimensional data. During each session, three image sets were taken, with 70, 1.5 mm

thick slices acquired continuously. Two sets had in-plane resolution of 0.49 x 0.49 mm; one acquired in the sagittal direction, and the second in the coronal direction. The third set was obtained in the sagittal direction with a resolution of 0.63 x 0.75 mm. Both the coronal and sagittal slices were acquired for each participant's knee joint allowing identification of discrepancies between the medial to lateral, anterior to posterior, and superior to inferior compartments of the knee joint (see Figures 6.3 and 6.4). The knee was scanned in a non weight-bearing position.

6.2.3. Osteoarthritis progression evaluation.

To evaluate the presence and progression of knee joint osteoarthritis, the T1-weighted scans were analysed in each plane of motion individually. This process was performed twice, on two days separated by one week, to assess the reliability of the results.

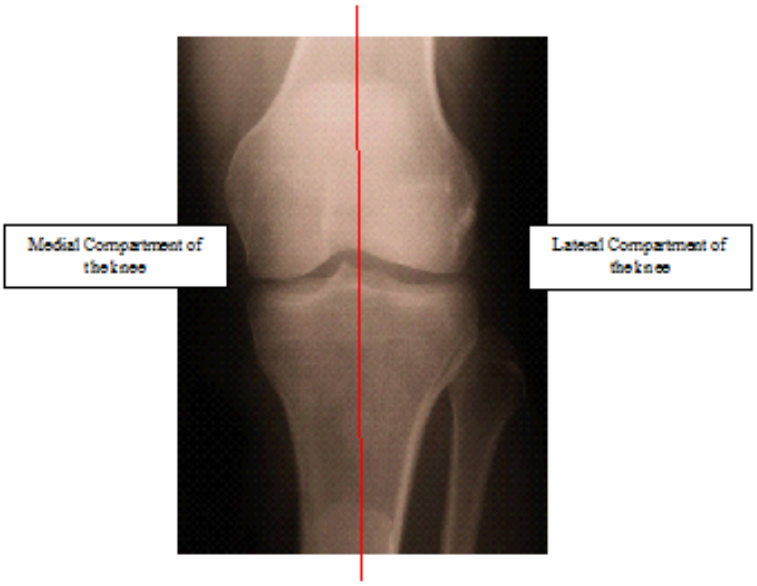


Figure 6.3. Deep anterior coronal view of the left knee joint illustrating the distinction of medial to lateral compartments of the femur and tibia. Picture (knee) taken from ClipArt, Office Word (2007).

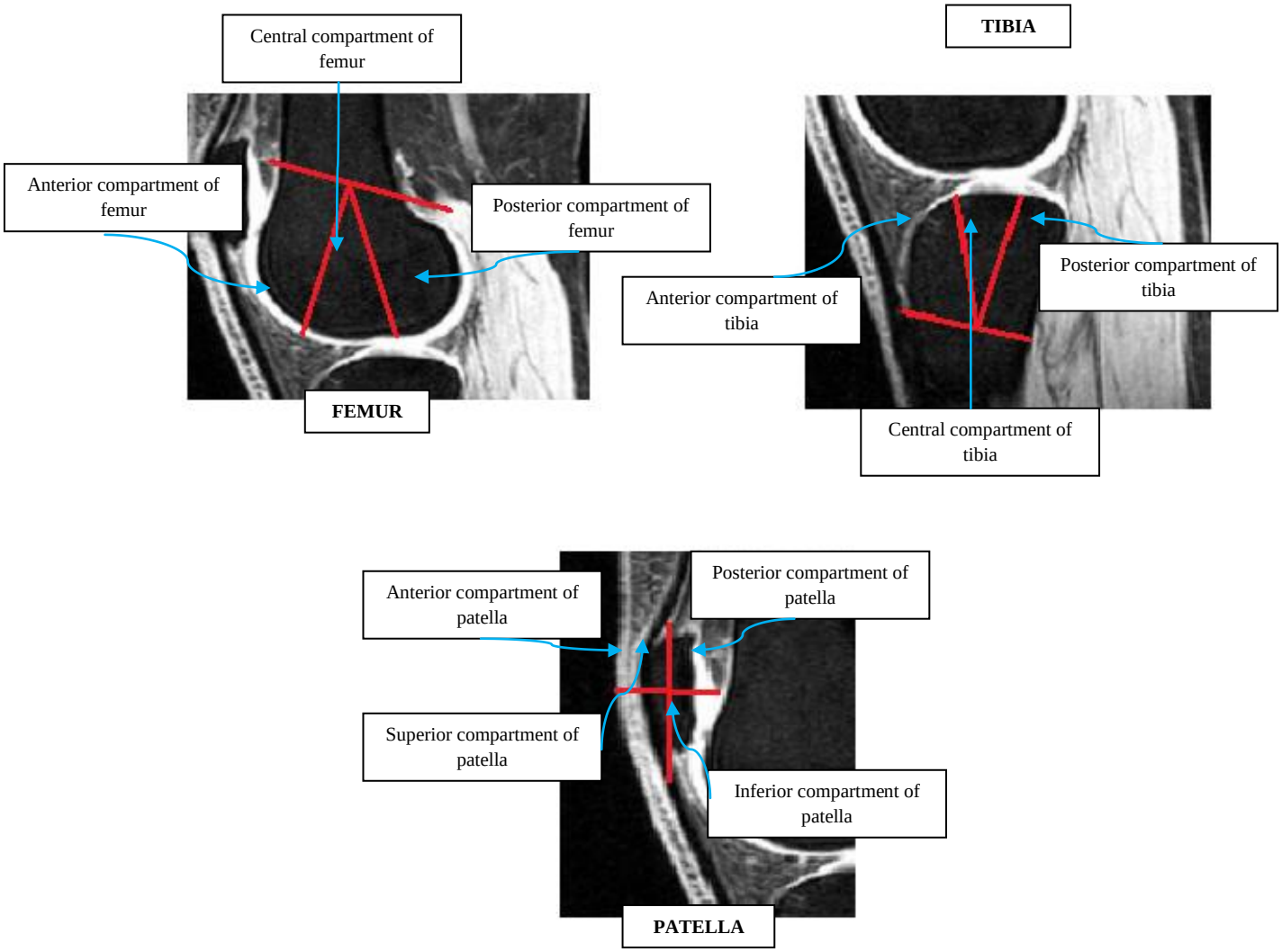


Figure 6.4. Sample MRI scan of the right knee in the sagittal plane. Diagram highlights the method of sectioning of the femur, tibia and patella, for osteoarthritis evaluation. MRI scans taken from the pilot study.

As illustrated in Figures 6.3 and 6.4, each femur and tibia was divided into medial, lateral and central compartments, and each patella was divided into four quadrants, resulting in ten potential sites of osteoarthritis development. Early stage osteoarthritis of the knee joint was identified through the presence of features associated with the condition (See Figure 6.2). Within the different compartments of the knee joint, osteophytes, bone oedema, enlarged tibial spines and sub articular cysts were identified and the exact location documented. These features were distinguished with the help of a local orthopaedic consultant, who was involved in the analysis of data. Each slice taken from every set of scans was scrutinised three times to aid reliability of results and limit the possibility of overlooking small occluded features. This assessment was then repeated on a separate day a week later, and the two sets of data compared with a test-retest assessment of reliability.

Osteophytes were identified as small formations of bone protruding into the articular cartilage surrounding the surface. They were identified as small hooks, commonly found on outer edges of each bone analysed (Kraan & Berg, 2007). The bone oedema was identified as ill-defined areas of intermediate to low signal on the T1-weighted images, presenting as white patches on the black bone, indicating increased fluid in that specific area (Pessis et al., 2003). The beaked tibial spines were identified in the coronal view, with the appearance of sharp hooks on each spine. Finally, the sub-articular cysts were associated with small circular areas of definitive white, appearing in the deep surface of the bone.

The severity of osteoarthritis present in each knee joint was quantified based on the number of features identified, and the existence with regard to the progression of osteoarthritis. The presence of osteophytes, bone oedma, beaked tibial spines, and sub articular cysts were graded with a score of one point for each single feature, and two points for each cyst as these are a later feature of osteoarthritis (Amin et al., 2005). As such, each knee was allocated a final score; the greater the value, the increased presence of early stage osteoarthritis. Medial and lateral osteoarthritis were categorised separately.

6.2.4. Correlation tests.

6.2.4.1. Correlation of MRI data with Knee Osteoarthritis Observation Survey.

A knee Osteoarthritis Observation Survey (KOOS) was completed by each participant prior to involvement in the biomechanical running investigation. This survey assessed the presence and potential for running injuries and osteoarthritis at the knee joint, within the sub categories of pain, symptoms, activities of daily living, sports and recreation, and quality of life. Each answer categorised osteoarthritis on a Likert Scale 0-5. Results indicated the severity of symptomatic osteoarthritis, with an average score of 100 representing a symptom free participant. To enable a correlation test, results were inverted, and the KOOS score was taken from 100 to give a positive value of osteoarthritis present. Participant mean results were then directly correlated with the results from the scans, to assess the relationship between symptomatic and MRI identified osteoarthritis at the knee joint.

6.2.4.2. Correlation of MRI data with age and menopause.

The quantified MRI data was initially correlated with the age of each participant, to evaluate the trend of increased severity and age among the female population. A simple correlation test was performed with the two sets of data, and a follow up regression analysis took place to assess the statistical significance of the r value. The correlation coefficient was deemed significant at the $p<0.05$ level. The results from the MRI scanning session were also correlated with information attained regarding the menopause and the supplementation of hormones. This was performed to determine whether any relationship existed between osteoarthritis and hormone replacement therapy, specifically oestrogen enhancement.

6.2.4.3. Correlation of MRI data with biomechanical data.

The biomechanical data involved in this study was taken from the biomechanical motion analysis investigation (Study 3) that had occurred two weeks in advance. The peak knee external adductor moment produced during running trials in a neutral running shoe was collected for each mature female who underwent MRI scans of the knee joint. The moment was calculated using inverse dynamics methods described in the previous chapters. The moment was only taken for the right knee, and was then correlated with the results from the MRI scans. The correlation test was performed with the grade allocated to each knee (medial and lateral osteoarthritis distinguished), and the average peak knee external adductor moment as calculated from ten running trials. A regression analysis then occurred to obtain the statistical significance of the correlation.

6.3. Results.

6.3.1. Participant information.

Ten asymptomatic mature female runners, each with one knee studied, met the inclusion criteria and volunteered for participation in this study. Table 6.2 presents participant information, including details of age, current menstruation, and any additional hormone supplements.

Table 6.2. Table displaying participant information, including date of birth, years of menopause, and the hormone replacement therapy.

PARTICIPANT	DOB/AGE	MENOPAUSE	OESTROGEN SUPPLIMENT
1	12/11/64: 46 years	Irregular 18 months.	No
2	03/04/1952: 59 years	>5 years	No
3	23/04/1967: 44 years	N/A	N/A
4	14/05/59: 52 years	>5 years	No
5	18/07/62: 49 years	2 years	No
6	06/01/58: 53 years	>5 years	No
7	24/08/57: 54 years	3 years	No
8	13/12/1953: 58 years	> 5 years	No
9	23/02/64: 47 years	N/A	N/A
10	18/10/54: 56 years	> 5 years	No
MEAN	51.8		
STANDARD DEVIATION	5.15		

6.3.2. Results from the MRI scans.

Of the female participants that volunteered for participation in this study, eight showed signs of early stage osteoarthritis at the knee joint. One set of scans showed the presence of metal artefact, which occluded a substantial proportion of the knee. This knee was therefore not included in the analyses. One female was classified as symptom free (Participant 3), with no signs of osteoarthritis present in any compartment of the knee joint. This knee was still incorporated in the analyses, as its inclusion was required for the correlation test.

A set of sample images are displayed to illustrate some of the features of osteoarthritis described in the following results tables. Figures 6.5a-b and 6.6 illustrate sample scans acquired in the sagittal plane. Figure 6.5a highlights the presence of an osteophyte on the posterior tibia, compared with a knee showing no signs of osteoarthritis (Figure 6.5b). The presence of bone oedema on the posterior patella is highlighted in Figure 6.6. This was identified as the area of white activity on the bone, and was measured at 6 mm.

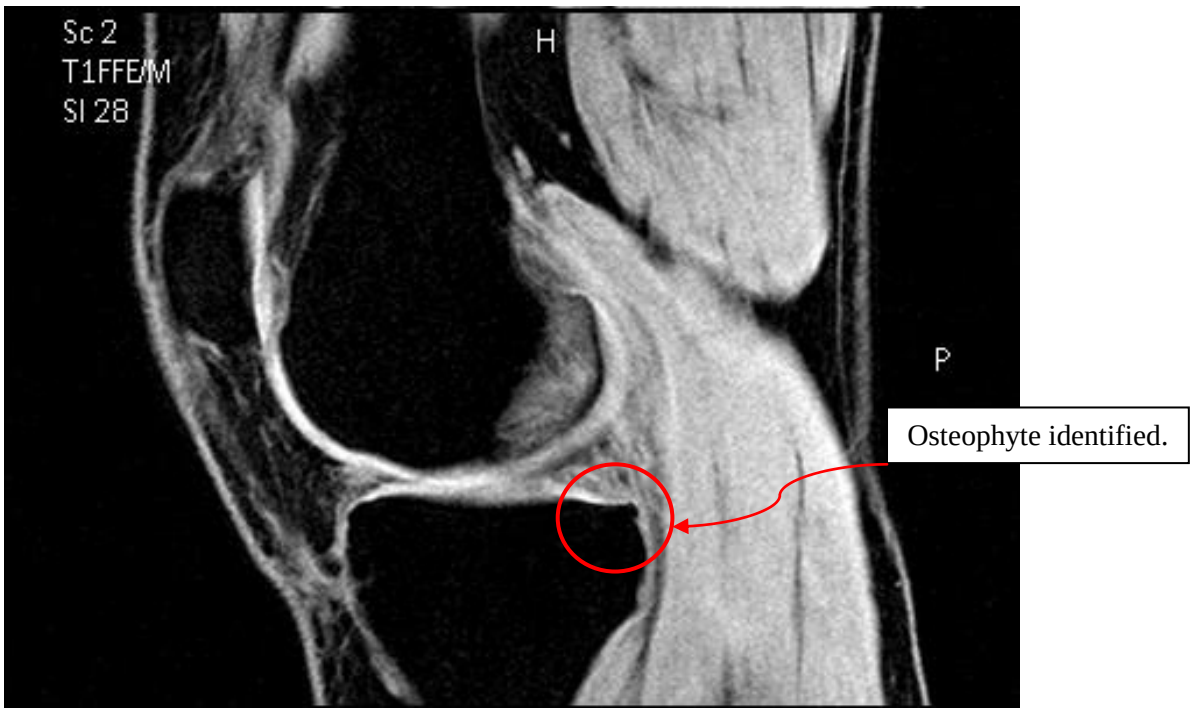


Figure 6.5a. Sample slice (28) acquired in the sagittal view illustrating the femur, tibia and patella. Osteophyte highlighted on the posterior tibia (Participant 1).

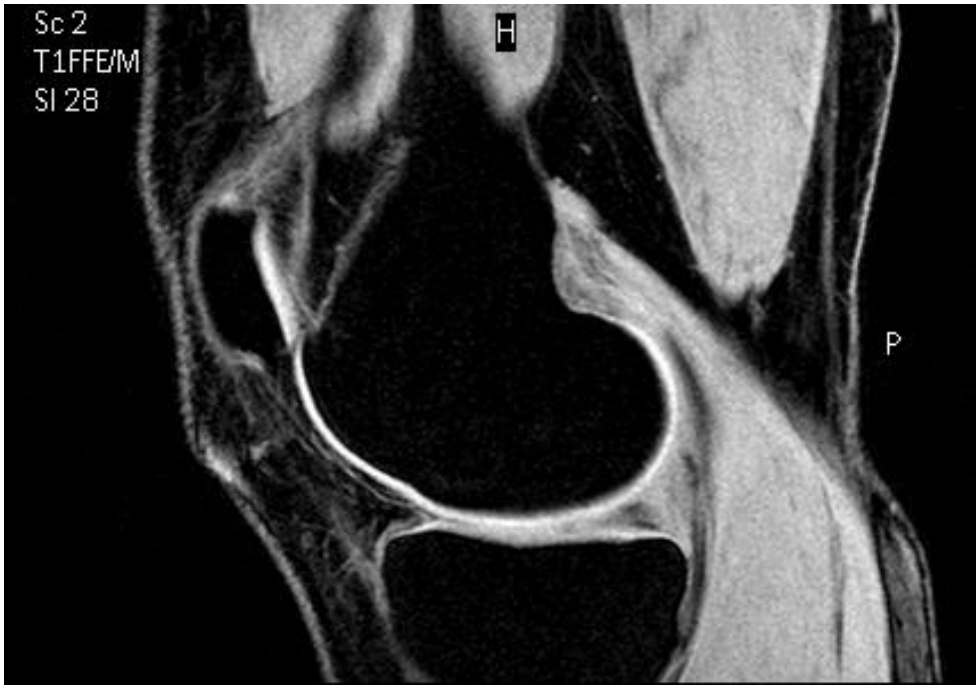


Figure 6.5b. Sample slice (28) acquired in the sagittal view illustrating the femur, tibia and patella. No features were identified (Participant 3).

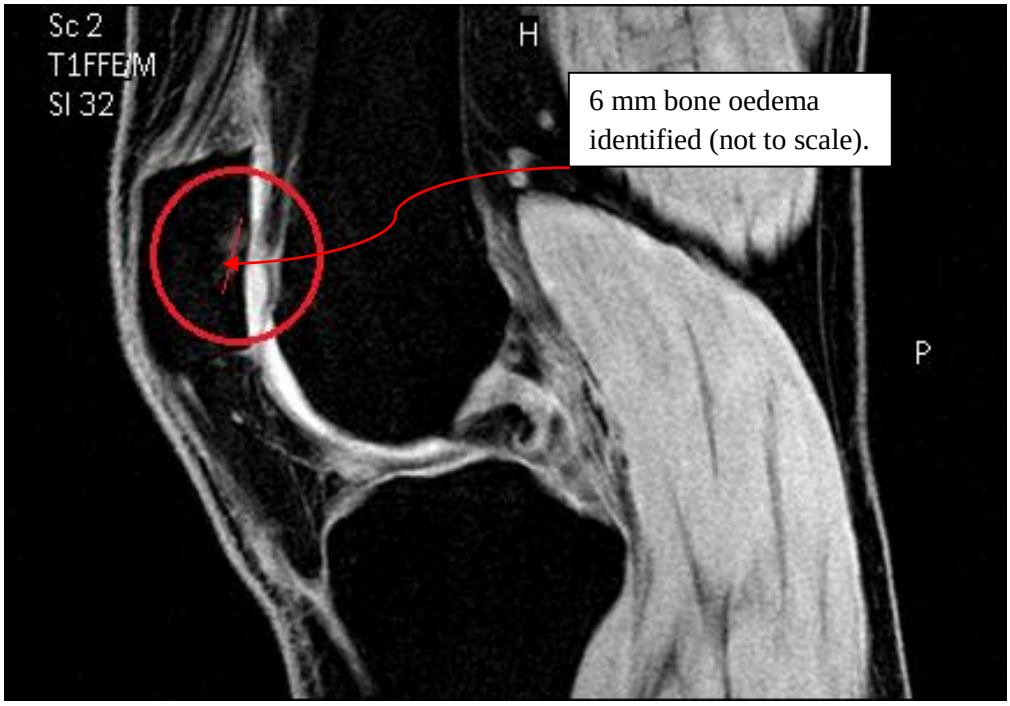


Figure 6.6. Slice acquired in the sagittal view highlighting bone oedema on the posterior patella (Participant 1).

Assessment of the results was initially qualitative, with signs of osteoarthritis determined from the presence of osteophytes, bone oedema, and subarticular cysts. Tables 6.3a-c illustrates the qualitative results from the MRI scans of the knees of each female. As highlighted, osteophytes were the most common feature identified, with all eight knees that showed signs of osteoarthritis, presenting a minimum of one (Table 6.3a). Bone oedema and beaked tibial spines appeared less frequently, both presenting in only 37.5% of the knees (Table 6.3b and c). Sub-articular cysts, a later feature of osteoarthritis, were only identified in one knee (Table 6.3c). Assessment of the knee joint in the coronal plane illustrated discrepancies in osteoarthritis between the medial and lateral compartments of the femur and tibia (Figure 6.3). Six knees showed signs of early osteoarthritis in the femur and tibia, with 79% of the features identified, isolated to the medial compartment.

Tables 6.3a-c. Tables illustrating qualitative results of the MRI scans, with the location of osteophytes (a), bone oedema (b), beaked tibial spines and (c) subarticular cysts identified for each female.

a)

PARTICIPANT	**OSTEOPHYTES**					
	PATELLA		**FEMUR**		**TIBIA**	
	Inferior	Superior	Medial	Lateral	Medial	Lateral
1	Osteophyte	Osteophyte	*No features identified.*	*No features identified.*	Osteophyte on medial tibial plateau. Osteophyte on medial tibial spine.	*No features identified.*
2	Osteophyte	Osteophyte	Osteophyte on patella-femoral area		Osteophyte on medial tibial plateau.	*No features identified.*
3	*No features identified.*	*No features identified.*	*No features identified.*			
4	Osteophyte	Osteophyte	Osteophyte on medial intercondyle notch. Osteophyte on medial femoral condyle.	Osteophyte on lateral intercondyle notch.	Osteophyte on central posterior tibia.	
5	*No features identified.*	Osteophyte	*No features identified.*			
6	Osteophyte	Osteophyte	Osteophyte on anterior surface of femur.		Osteophyte on posterior medial tibia	Osteophyte on anterior lateral tibia.
			Osteophyte on medial femoral condyle.	Osteophyte on lateral femoral condyle.		
7	Osteophyte	Osteophyte	Osteophyte on anterior patella-femoral surface.	*No features identified.*	Osteophyte on medial posterior tibia.	*No features identified.*
8	*No features identified.*	Osteophyte	*No features identified.*		Osteophyte on anterior tibia.	*No features identified.*
9	LARGE AMOUNT OF METAL ARTIFACT OCCLUDED VISION.					
10	Osteophyte	Osteophyte	*No features identified.*			

(b)

PARTICIPANT	**BONE OEDEMA**					
	PATELLA		FEMUR		TIBIA	
	Inferior	Superior	Medial	Lateral	Medial	Lateral
1	6 mm at midpoint of inferior surface.	*No features identified.*	7.9 mm on posterior third of medial femur.	*No features identified.*		
2	*No features identified.*					
3	*No features identified.*					
4	*No features identified.*					
5	*No features identified.*					
6	*No features identified.*	4.4 mm on superior patella on deep surface.	*No features identified.*			
7	*No features identified.*					
8	*No features identified.*					
9	LARGE AMOUNT OF METAL ARTIFACT OCCLUDED VISION.					
10	*No features identified.*	6 mm on superior patella.	*No features identified.*			

(c)

PARTICIPANT	**TIBIAL SPINES**	**SUB-ARTICULAR CYSTS**	
	TIBIA	PATELLA	
		Superior	Inferior
1	Slight beaking due to osteophyte presence	*No features identified.*	
2	*No features identified.*		
3	*No features identified.*		
4	*No features identified.*		
5	*No features identified.*		
6	Beaking of tibial spines	*No features identified.*	
7	Beaking of tibial spines	*No features identified.*	
8	*No features identified.*		
9	LARGE AMOUNT OF METAL ARTIFACT OCCLUDED VISION.		
10	*No features identified.*	2 Sub articular cysts present on superior patella.	*No features identified.*

The qualitative results as presented in the above tables (Table 6.3a-c) were quantified using a scale as described in the methods section. The results from the biomechanical assessment of knee external adductor moment during running are presented in Table 6.4, alongside the quantified data from the osteoarthritis evaluation. A sample moment time history for the knee external adductor moment is presented in Figure 6.7 (Subject 2).

Table 6.4. Table displaying quantified osteoarthritis results alongside previously obtained knee external adductor moments.

PARTICIPANT	KNEE OA SCORE	KEAM (Nm/kg)	KOOS RESULTS (100 – Score)
1	7	1.27	85.35 = 14.65
2	4	1.03	93.79 = 6.21
3	0	0.89	94.14 = 5.86
4	6	1.17	92.225 = 7.775
5	1	1.10	98.15 = 1.85
6	9	1.37	89.84 = 10.16
7	5	1.19	98.91 = 1.09
8	2	1.04	94.66 = 5.34
10	7	1.24	83.27 = 16.73
MEAN	**4.56**	**1.14**	**92.25/7.74**
STANDARD DEVIATION	**3.05**	**0.15**	**4.99**

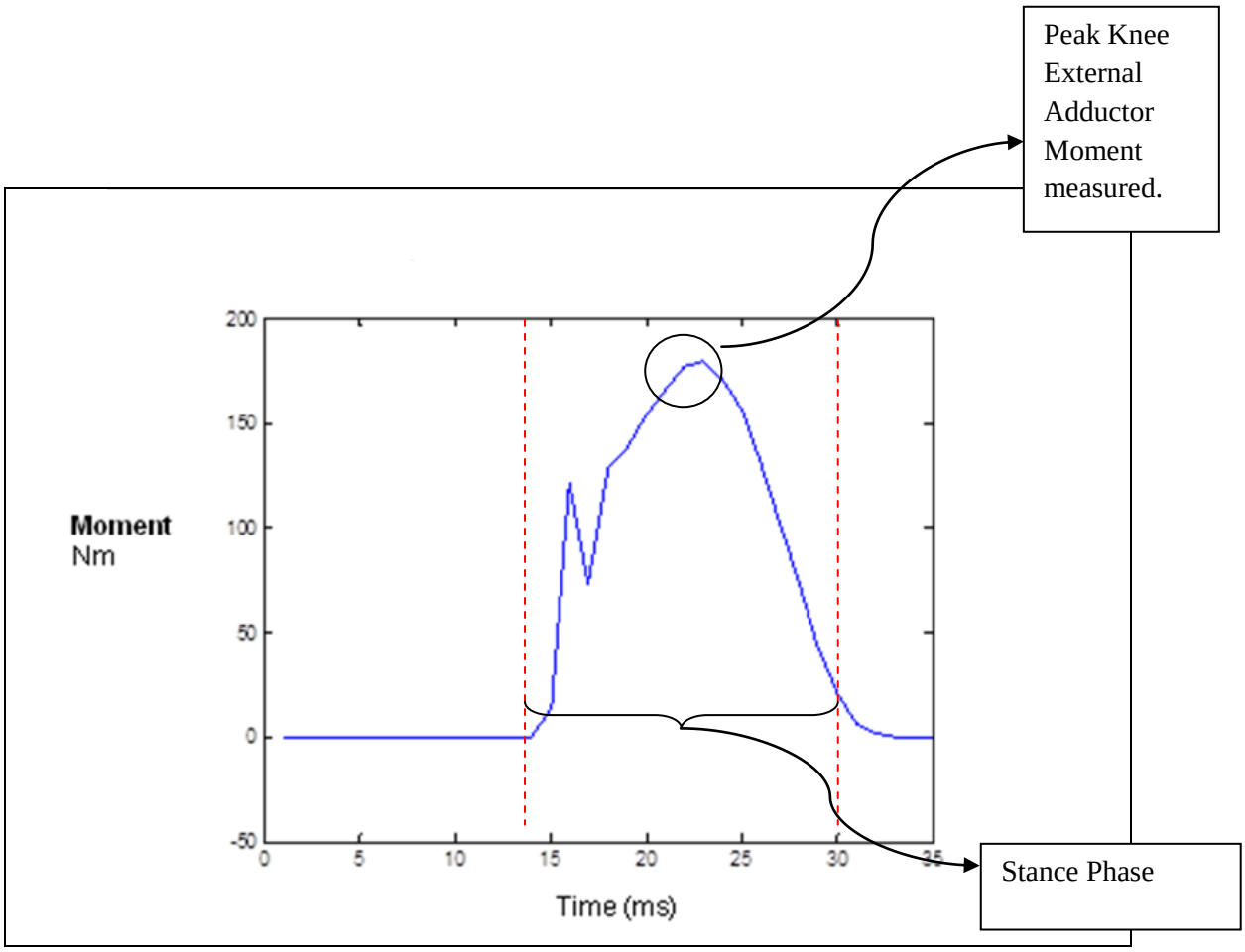


Figure 6.7. Sample graph illustrating the knee external adductor moment produced during one running stride. Stance phase and measured peak are highlighted.

6.3.3. Correlation of MRI score with age, KOOS score, and biomechanical data.

Correlation tests were performed to assess the association of knee osteoarthritis score with age, KOOS results and the knee external adductor moment (Figures 6.8 a-c). Table 6.5 illustrates the results of the follow up regression analysis. The correlation between knee osteoarthritis and age (0.22) was not significant ($p>0.05$), although when presented graphically, a moderate positive linear relationship is shown by the trend line (Figure 6.8a). The relationship between knee osteoarthritis and KOOS score showed a positive correlation coefficient ($r=0.64$) although this too was not statistically significant ($p>0.05$) (Figure 6.8b). However, similar to the osteoarthritis and age relationship, Figure 6.8b shows the trend line to demonstrate a positive linear relationship between the two variables.

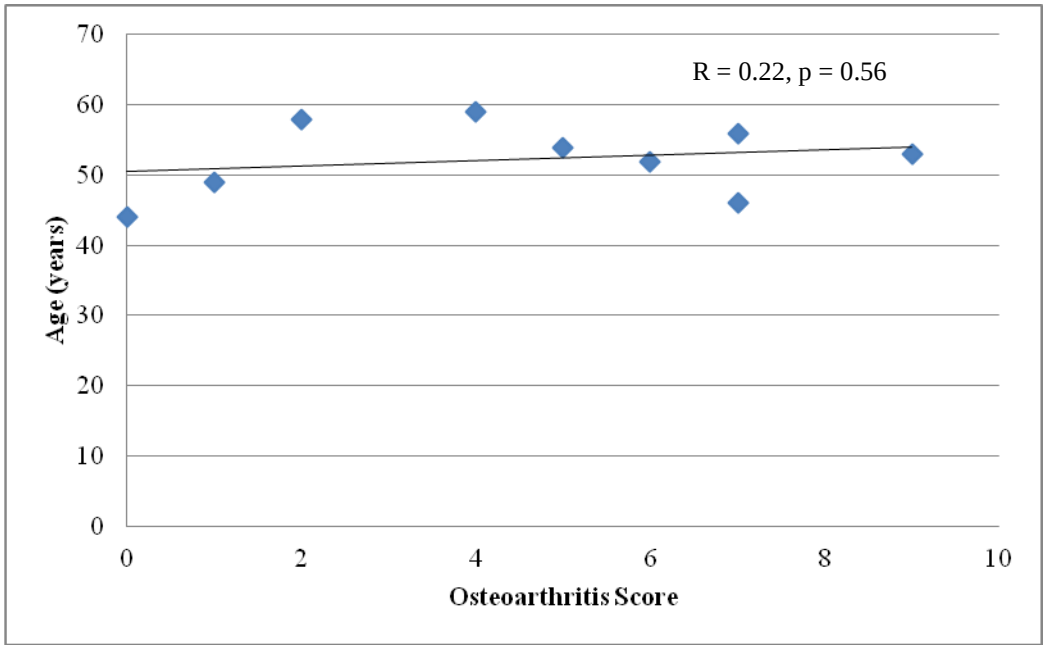
A correlation and regression test indicated a significant positive correlation between the osteoarthritis outcome and the knee external adductor moment among the mature female runners ($p<0.05$) (Table 6.5). Furthermore, the regression analysis demonstrated an R-square value of 0.83, suggesting that the regression line (Figure 6.8c) approximates 83% of the data. A scatter diagram highlights the association between knee external adductor moment and osteoarthritis presence, with the best fit line showing a linear correlation (Figure 6.8c).

Table 6.5. Results from correlation coefficient test and regression analysis tests for MRI results with age, KOOS results, and the knee external adductor moment (KEAM).

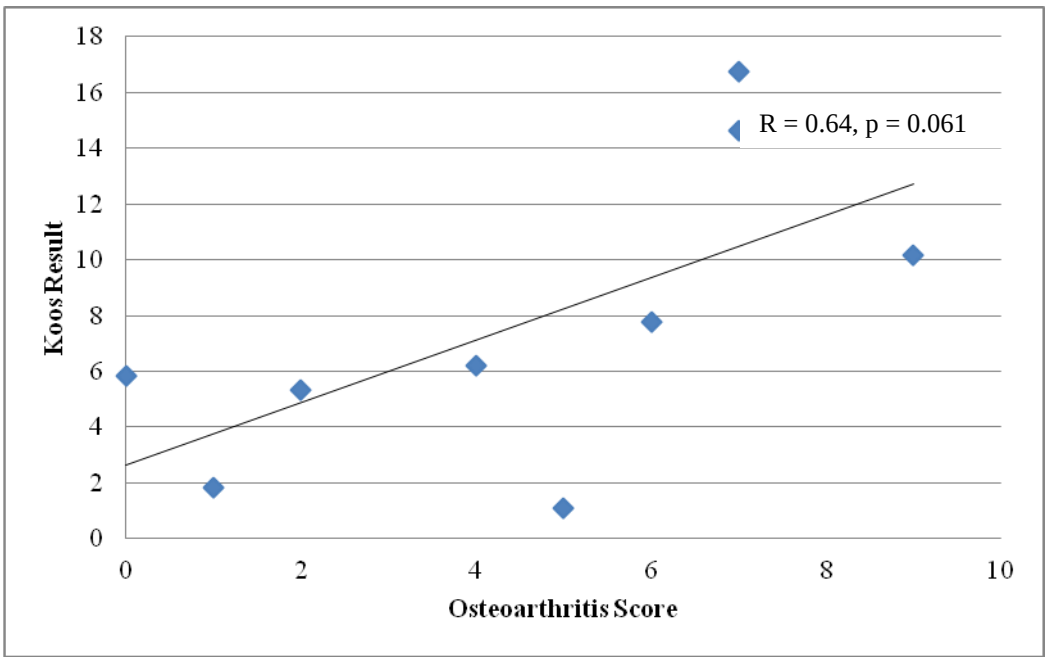
STATISTICAL ANALYSIS OUTPUT	MRI:AGE	MRI:KOOS	MRI:KEAM
Correlation Coefficient	0.22	0.64	0.92
R Square	0.05	0.41	0.83
Observations	9	9	9
P-Value (2 sf)	0.56	0.061	**0.000512***

Significant difference highlighted (*).

(a)


R = 0.22, p = 0.56
Age (years)
70
60
50
40
30
20
10
0
0
2
4
6
8
10
Osteoarthritis Score

(b)


R = 0.64, p = 0.061
Koos Result
18
16
14
12
10
8
6
4
2
0
0
2
4
6
8
10
Osteoarthritis Score

(c)

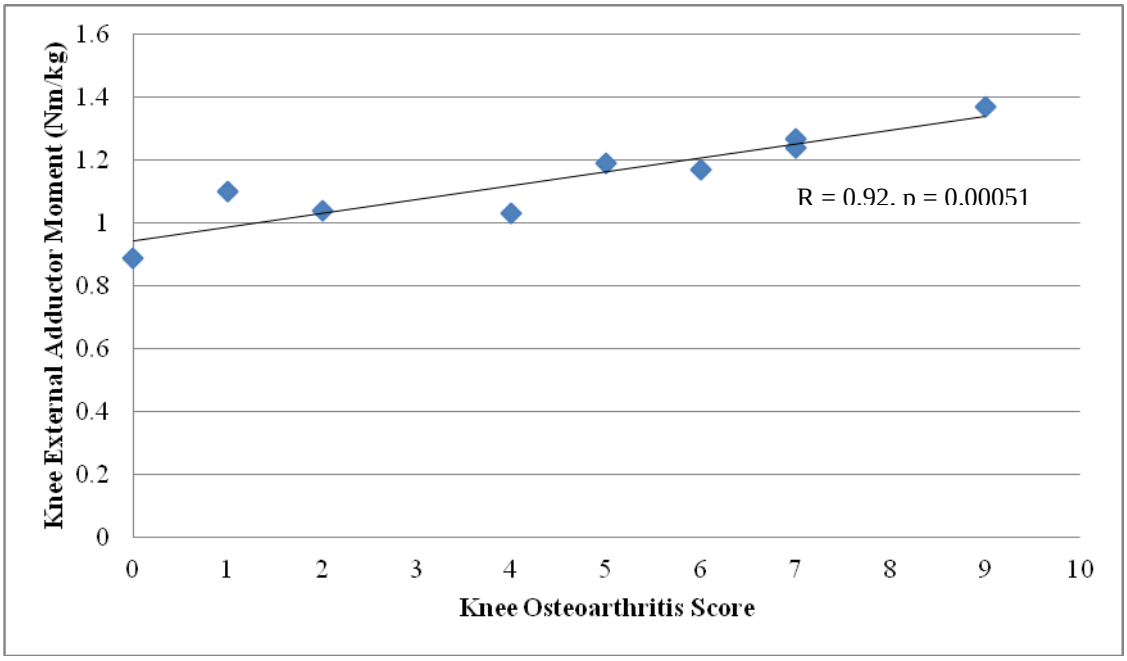


Figure 6.8 a-c. Scatter diagrams illustrating the relationship between osteoarthritis present at the knee joint and age (a), KOOS score (b) and the knee external adductor moment (c) among nine mature females.

6.3.4. Reliability test.

Results from the test-retest of reliability illustrated a correlation between the MRI results assessed on two separate days (Figure 6.9). As the more subtle of features, the test-retest was performed with the results from osteophyte identification. The results showed a perfect positive correlation of 1.0 ($p<0.01$).

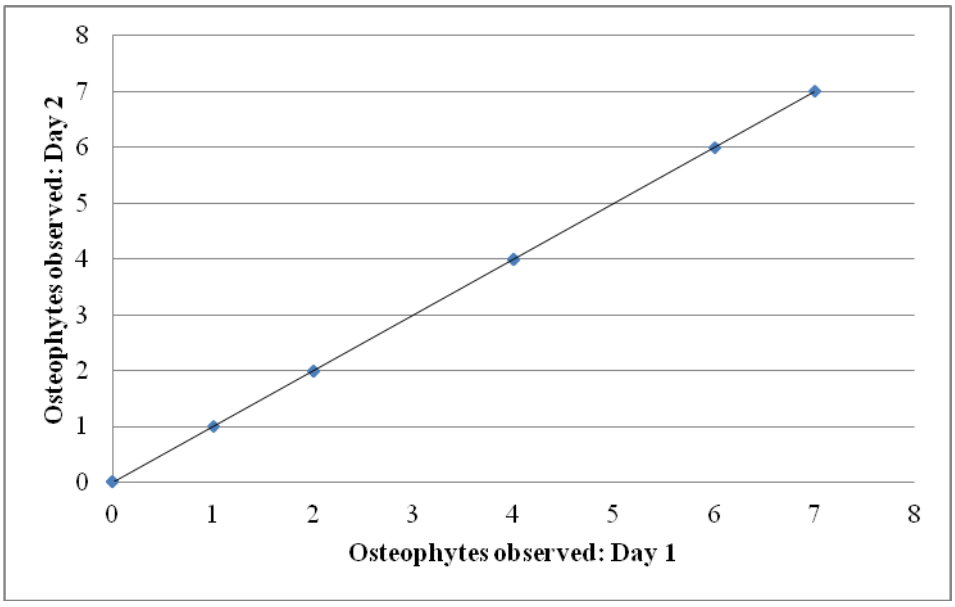


Figure 6.9. Correlation of MRI scores produced on day 1 compared with a second assessment on day 2.

6.4. Discussion.

The aim of this study support the original theory that mature female runners are at a higher risk for the development of debilitating conditions, specifically knee joint osteoarthritis. The purpose was to assess the current condition of the knee joint of each participant, and investigate the relationship with age, KOOS results, and knee external adductor moment. Ten mature female runners of similar running ability (determined through results of the most recent 10 km race) volunteered to take part following involvement in the ongoing biomechanical assessment. Each female was classified as symptom free from the results of a completed knee osteoarthritis observation survey (KOOS, see chapter 5, section 5.3.1.1). Magnetic resonance imaging techniques were employed to identify changes in the joint, which are commonly unrecognisable through radiographic assessments. Results were then combined with those produced from the ongoing biomechanical investigation, and a correlation test was performed to assess the relationship between the knee external adductor moment and the scale of osteoarthritis present in the knee joint.

6.4.1. Knee joint osteoarthritis evaluation.

Of the ten female participants that underwent magnetic resonance imaging scans, eight displayed signs of early stage osteoarthritis; each diagnosed knee showing the formation of at least one osteophyte (Figure 6.5a). Osteophytes are fibrocartilage-capped bony outgrowths, which take the form of one of three types; the traction spur, the inflammatory spur, and the genuine osteophyte (Kraan, & Berg, 2007). The former two are located at the insertion of tendons and ligaments, with the inflammatory spur represented by the syndesmophyte; a bony growth formed inside a ligament (Menkes & Lane, 2003). The genuine osteophyte, commonly known as osteochondrophyte, presents at the periosteum overlying the bone at the junctional zone between cartilage and bone (Menkes and Lane, 2003). It is this latter form of osteophytes that were identified in the knee joints of the mature female runners.

The clinical relevance of osteophyte formation was summarised by Kraan and Berg (2007), who suggested that osteophytes are a main feature of osteoarthritis, and form in the early development of the condition, often identified prior to joint space narrowing. In support of this, Bullough (1992) stated that joints affected by osteoarthritis are characterised by the production of new connective tissue in the form of osteophytes. Osteophytes have also been identified in experimentally induced osteoarthritis of the knee joint, supporting this feature's role in the development of the condition (Moskowitz and Goldberg, 1987). The presence of osteophytes in the knees of mature females in the current study and the positive correlation with the knee external adductor moment support this suggestion; that osteophyte presence is a sign of early stage osteoarthritis.

However, the link between osteophyte formation and osteoarthritis is not equivocal, and several researchers have more clearly linked osteophyte presence with positive effects on distribution of forces. According to Menkes and Lane (2003), osteophyte formation is considered to be an adaptive reaction of the joint to cope with instability, and plays a compensatory role in the redistribution of forces, aiding articular cartilage protection. Similarly, Pottenger, Phillips and Draganich, (1990) demonstrated the removal of osteophytes to increase the varus-valgus movement at the knee joint, highlighting the role of these features in joint stabilisation. However, it is noted that these knees had previously been diagnosed with severe osteoarthritis, and the role of osteophytes as a cause or consequence of the condition is not clear. It is however apparent that, despite uncertainty regarding the order of appearance, osteophyte formation is succinct with osteoarthritis, and can therefore be considered a feature associated with the condition.

An additional feature identified in the knees of the mature females in the current study is the beaking of the tibial spines. According to Holten (2004) the beaked tibial spines are a positive finding in diagnosis of knee joint osteoarthritis, supporting the suggestion that the mature females were illustrating early stage osteoarthritis of the knee. This is further supported by a recent investigation, analysing the morphological features of osteoarthritis in paleopathological specimens. Hayeri and colleagues indicated a positive correlation between tibial spine height and grading of osteoarthritis, confirming it as a feature of the condition (Hayeri, Shiehmorteza, Trudell, Hefflin, & Resnick, 2010). Similarly, Hayeri et al., (2010) associated tibial spine beaking with the development of osteophytes in early

stage osteoarthritis. This association was supported in the results from the current study, as osteophytes were present in each knee that displayed beaked tibial spines.

Bone oedema was identified in three of the eight knees that showed signs of early stage osteoarthritis. This was identified as signs of increased activity producing areas of white colour on the scans of the bone (Figure 6.6). The identification of this feature not only supports the suggestion that the mature females were presenting with early stage knee osteoarthritis, but also supports the use of magnetic resonance imaging techniques in diagnosis, as oedema is often unidentifiable with radiography (McQueen & Ostendorf, 2006). The association of bone oedema as an early sign of osteoarthritis is supported by Felson and colleagues, who performed a longitudinal study into the progression of osteoarthritis of the knee joint. Here it was shown that risk of progression was increased six fold among patients with signs of bone oedema (Felson et la., 2001). This was further supported by Garnero and colleagues who associated the presence of bone oedema with increased collagen breakdown, and McQueen (2007) who suggested bone oedema to be intimately involved in the pathological process, predicting radiographic erosive progression (Garnero, Peterfy, Zaim, & Schoenharting, 2005; McQueen, 2007). Both investigations therefore lend support to the findings and assumptions that the bone oedema identified among the mature females is a sign of osteoarthritis.

The presence of early stage knee joint osteoarthritis in the knee joint of one participant was supported by the identification of subarticular cysts. Bone cysts have been well recognised as features of advanced osteoarthritis for decades, often taking one of two forms (Rhaney & Lamb, 1955; Crema, Roemer, Marra, Niu, Lynch, Felson, & Guermaz, 2010). The synovial fluid intrusion theory suggests that increases in sub-articular pressure forces fluid into the subchondral bone via damaged cartilage causing cysts to form deep within areas of eroded cartilage (Landells, 1953). The second form of cyst, known as the bony contusion cyst, is a consequence of traumatic impact between two opposing articular surfaces (Crema et al., 2010). As a loss of cartilage was not reported for any participant within this investigation, the subchondral cysts identified in the knee of participant 10 are likely be of the latter form. This therefore suggests that osteoarthritis is present in this knee due to uneven articulation between the patella and femur.

6.4.2. Osteoarthritis among asymptomatic mature female runners.

6.4.2.1. Osteoarthritis among mature members of the female population.

As all participants were female, with an average age of 51.8 (±5.15) years, each runner was assumed to be in a high risk category for the development of osteoarthritis at the knee joint (Moskowitz et al., 2007). Despite this, throughout the previous chapters and investigations, all mature female runners were characterised as free from symptoms of debilitating knee conditions. Results from the magnetic resonance imaging scans illustrated that out of the nine knees included in the analysis, eight presented with early stage knee osteoarthritis. This alone supports the previously presented theory that there is a poor association between symptomatic and diagnosed osteoarthritis at the knee joint (Hannan et al., 2000).

Although males were not included in this study for comparative purposes, the presence of osteoarthritis identified in 89 % of the females could be considered to support the theory that osteoarthritis is common in women. Results from the Framingham Osteoarthritis Study (Felson, 1990), showed women to present an overall higher level of radiographic osteoarthritis compared to a group of aged matched men. However, as suggested, no such comparison occurred in the current study, and the results can only be used to make inferences on the population of mature females based on a sample of 9 knees.

6.4.2.2. Osteoarthritis among runners.

The relationship of physical activity to the development and progression of the condition has previously been examined with mixed results. Our investigation included ten participants who were classified as recreational runners, of which eight presented with early stage osteoarthritis. As no comparison was made with a control group of sedentary aged-matched females, these results enable only speculation on the role of exercise in osteoarthritis development.

An early case control study performed by Imeokparia and colleagues, suggested that the female gender, physical activity levels, and age were all risk factors for the development of osteoarthritis at the knee joint (Imeokparia, Barrett, Arrieta, Leaverton, Wilson Hall and Marlowe, 1994). This association between osteoarthritis development and physical activity was further supported in an additional study which looked at the development of the condition among ninety two participants. Although this latter study displayed a significant effect for physical activity, information regarding osteoarthritis presence was determined through questionnaires (Kohatsu and Schurman, 1990).

An additional study shown to support the association between physical activity and osteoarthritis was performed with dancers (Teitz and Kilcoyne, 1998). These results indicated the level and frequency of dancing undertaken to significantly increase the risk of premature symptomatic and radiographic osteoarthritis. Additionally, a prospective study performed with the Framingham study data indicated that heavy levels of physical activity undertaken by the elderly was a risk factor for osteoarthritis development, independent of previous injuries or participation in elite sports. However, neither investigation directly looked at the effect of long distance running.

In contrast, a study performed at a similar time suggested that long distance running, even at elite level, may not be an independent risk factor for the development of knee osteoarthritis (Kujala, Kettunen, and Paananen, 1995). Similarly, a more recent longitudinal study found that running was not associated with accelerated incidence or radiographic progression of knee joint osteoarthritis (Chakravarty, Hubert, Lingala, Zatarain, and Fries, 2008). It is important to note here however, that when compared to a group of inactive participants, the long distance runners did have increased radiographic osteoarthritis at baseline.

As the results from our study showed 88% of runners to display early signs of osteoarthritis, with no comparison of a control group, it is uncertain whether the activity of running is an independent risk factor for this incidence. It is possible that an aged-matched group of sedentary females may present similar signs of osteoarthritis. As such, future longitudinal research is needed to further understand the relationship between running and osteoarthritis, with assessments among a group of gender and aged matched individuals, free from the condition at baseline.

6.4.2.3. Osteoarthritis and age.

Although all females were classed as mature based on inclusion in the 40 – 60 years category, the participants ranged in age from 44 years to 59 years. As such, a correlation test was performed to assess the effect that age had on the scale of osteoarthritis identified at the knee joint. As illustrated in Table 6.5, although a moderate linear relationship was present between the two variables (Figure 6.8a), the correlation coefficient was not statistically significant ($p>0.05$).

This result is in contrast to theories presented in the literature, suggesting that increasing age is a risk factor for osteoarthritis development and progression (Felson, 1990). It has been commonly suggested that age is directly proportional to risk of osteoarthritis, and according to Loeser (2000), by the age of 60 years, 100% of the population will have signs of osteoarthritis at one or more joints. Similarly, in the Framingham osteoarthritis study (age range 63-94 years), knee osteoarthritis was seen to progressively increase with age, with a rate of 4% per year among women (Felson, Naimark, Anderson, Kazis, Castelli & Meenan, 1987). Furthermore, a later study showed the prevalence of osteoarthritis to significantly increase with age in women but not in men, supporting the earlier association between osteoarthritis and gender (Felson, 1990).

Although the age range of the mature females was only 15 years, it was noted that there was a discrepancy in hormones, based on the premise that 33% of the group had not been through the menopause (Table 6.2). During the menopause, oestrogen is no longer cyclically released into the bloodstream, and as such many women are prescribed oestrogen replacement supplements. The information collected from the mature females however show that no participant was or had ever taken any form of hormone replacement therapy. As such, it was assumed that the three females that had not been through the menopause had higher levels of oestrogen than the post-menopausal group. Results from the MRI scans showed participant 1 (pre-menopausal) to have 7 signs of osteoarthritis at the knee joint (Table 6.4). Participant 3 however was also described as pre-menopausal, and was diagnosed as free from features of early stage osteoarthritis. It is also noted that this was the youngest female in the group. Participant 9 was described as pre-menopausal, although due to the presence of metal artefact, the association with osteoarthritis could not be analysed.

These results are therefore conflicting. However, while no direct relationship between hormonal status and osteoarthritis presence could be established within the scope of this study, the possibility of oestrogen levels influencing the results must be acknowledged. Within the literature, the effect of the menopause, and subsequent reduced levels of oestrogen on osteoarthritis is conflicting. Epidemiological studies have often provided support for an association between oestrogen deprivation and osteoarthritis development, by showing an increase in prevalence of the condition among women over the age of 50 years (Felson and Zhang, 1998). Prior to 50 years, incidence of osteoarthritis appeared greater among men compared to women, however beyond this age the prevalence was reversed (Felson and Zhang, 1998). Similarly, a positive relationship has been shown between oestrogen replacement therapy and reduced osteoarthritis risk, indicating a possible therapeutic role of oestrogen in osteoarthritis (Felson and Nevitt, 1998). This appears to support the general theory presented in the literature, although a small number of studies have indicated non-significant effects of hormone replacement therapy. One study showed no effect of hormone replacement therapy on osteoarthritis; however it is noted that the condition was monitored in the finger joints, and methodological differences limit the ability to directly compare with the results among the mature females in the current study (Maheu, Dreiser, Guillou, & Dewailly, 2000).

As the seven mature female runners that were categorised as post-menopausal with no hormonal supplements all showed signs of osteoarthritis, it is possible that the lack of oestrogen within the blood stream could be an additional factor contributing to the development of this condition at the knee joint. Although the overall effect of age on osteoarthritis scale among the mature females was not significant, the age range of the group was only 15 years, and the results were based on the osteoarthritis findings of only nine females. Therefore, it is proposed that an increase in sample size may have strengthened the linear relationship seen.

6.4.3. KOOS scores, osteoarthritis and biomechanical alignment.

6.4.3.1. Correlation between MRI evidence and KOOS scores.

Although pain is one of the most common symptoms of knee osteoarthritis, the non significant correlation between the KOOS survey and osteoarthritis shown among the mature females in the current study ($r = 0.64$, $p > 0.05$) support the theory that there is often a poor association between symptoms of pain and the early stages of the condition. This is similar to the results found by Barker and colleagues, who demonstrated a poor correlation between radiographic signs of early stage osteoarthritis and results from the WOMAC (Western Ontario and McMaster Universities) index ($p>0.05$) (Barker, Lamb, Toye, Jackson & Barrington, 2004). Similarly, Spector and Hart (1992) found less than 50 % of people presenting with radiographic signs of osteoarthritis to have symptoms related to the findings, demonstrating the poor relationship between self reported knee pain and severity of diagnosed osteoarthritis.

In contrast, Kornaat and colleagues found a strong association between osteophytes present in the patellofemoral area and pain reported among a group of osteoarthritis patients (Kornaat, Bloem, Cleulemans, Riyazi, Rosendaal, Nelissen, Carter, Le Graverand & Kloppenburg, 2006). It is however noted that all patients were classified with moderate to severe osteoarthritis, which may have contributed to the pain recorded.

The insignificant relationship between MRI evidence of knee osteoarthritis and symptoms could be related to the lack of cartilage degradation and joint space narrowing identified at the joints of the female runners at this stage. According to Ayral and colleagues, cartilage loss on arthroscopy had a strong positive correlation with knee pain and disability (Ayral, Dougados, Listrat, Bonvarlet, Simonnet, & Amor, 1996). However, as there are no pain receptors within articular cartilage, the association between cartilage loss and pain severity is ambiguous (Hunter, March, & Sambrook, 2003) and the lack of cartilage loss appears to have little relevance on the non significant correlation between osteoarthritis and pain among mature females in the current study.

Although not significant the correlation ($r = 0.64$) reported among the mature females suggests that a larger sample size in this investigation may have increased the

significance of the association between pain and osteoarthritis presence. According to Stephens, Martire, Cremeans-Smith, Druley, & Wojno, (2006) pain is a personal experience, and therefore the methods use to assess pain reliably among a variety of individuals is difficult. Therefore it may be that differences in inter-individual perception of pain among the females could be a limiting factor in the relationship between symptoms at osteoarthritis.

6.4.3.2. Correlation between KOOS and knee external adductor moment.

The relationship between KOOS results and the knee external adductor moment is explored in detail in Chapter 5. Here it is suggested that a positive correlation exists between the knee external adductor moment and symptoms at the knee joint, a result similar to that presented in the literature. Hurwitz and colleagues showed a significant positive correlation between symptoms of pain, and the knee external adductor moment in symptomatic osteoarthritis patients (Hurwitz, Ryals, Block, Sharma, Schnitzer & Andriacchi, 2000). It is important to note here that this previous study involved the use of symptomatic patients. Additionally, although a positive correlation did exist between KOOS scores and knee external adductor moment in the current study, all mature females were classified as symptom free. These results support the theory that the knee external adductor moment may linked with the development of early stage osteoarthritis.

It is important to note that although correlations with KOOS scores were demonstrated in the current study, results from this survey categorised all females as symptom free. Therefore, although variance was seen in the results produced by the mature females, the range of scores was small (83.27-98.15), basing correlation results on a small range of values. It is suggested that a larger sample size of mature female runners, including both symptomatic and asymptomatic participants, may strengthen the associations between osteoarthritis, biomechanical alignment and symptoms at the knee joint.

6.4.4. Relationship between MRI evidence of osteoarthritis and knee external adductor moment.

As described, although the theories within the current literature are not equivocal, it appears that the features identified among the mature female runners in the current study are likely signs of early stage knee osteoarthritis. A correlation was therefore performed to assess the relationship between the level of osteoarthritis present at the knee joint, and the knee external adductor moment recorded during running. Overall, the results from this test lend strong support to the positive association between the knee external adductor moment and knee joint osteoarthritis development (Figure 6.8c).

Osteoarthritis is widely believed to be the consequence of local mechanical factors acting within the context of systemic susceptibility (Sharma et al., 2001). At the joints, certain site-specific factors govern how load is distributed between the medial and lateral compartments, and the overall ability of the structures to manage that load (Sharma et al., 2001). At the knee joint, alignment is a key determinant of load distribution, and the knee external adductor moment is a known measure of medial knee joint loading, and increased varus position of the knees (Zhao et al., 2007). As such it was anticipated that participants with greater knee external adductor moments recorded during running trials would display increased levels of knee joint osteoarthritis compared those who ran with lower moments.

The correlation was performed with nine participants, and showed a positive correlation of R squared 0.83 ($p<0.05$), indicating that the increased knee external adductor moment strongly predicted the osteoarthritis level within the joint, with 83 % of the variation in osteoarthritis signs being explained by the knee external adductor moment. This result lies in accordance with previous biomechanical studies suggesting that increased knee external adductor moments are associated with increased medial knee joint loading (Foroughi et al., 2009). During gait, in the normally aligned knee joint, the load is disproportionately transmitted to the medial compartment, which carries up to 70% (Hunt et al., 2001). When an external adductor moment acts at the joint, the knee moves to assume a varus position, and the medial knee carries up to 100% of the load. As such, it was expected that the increased moments would be associated with increased signs of wear on the medial compartment. Among the mature females that presented signs of

osteoarthritis, in the coronal plane 79% was identified on the medial compartment. This therefore further supports the association between varus alignment, medial compartment loading, and joint degradation.

The average knee external adductor moments produced by the mature females were recorded as high, compared with those produced by younger females in the literature (Pollard, Davis, & Hamill, 2004). Additionally, direct comparisons with younger female runners in previous studies demonstrated the statistical significance of this difference (Lilley et al., 2011; Lilley et al., 2011). The average knee external adductor moment was taken from the second, higher peak in the moment time history trace, occurring at midstance of a running stride (Figure 6.7). This peak occurs as the knee moves into a varus position with the body weight positioned over the foot, increasing the moment arm between the line of ground reaction force and the knee joint centre (Chapter 5, Figure 5.1). The high moments produced by the mature females cause a high load to be consistently focussed at the medial knee during each foot stride taken in a run, suggested to result in the degeneration of that compartment. This notion is supported in the findings from the magnetic resonance imaging scans.

Although the features identified among the mature females are all categorised as signs of early stage osteoarthritis, the statistically significant correlation result suggests an association between knee external adductor moment, and severity of the condition. This result is similar to those presented in the literature; when comparing the knee external adductor moment among both less and more severe osteoarthritis patients as well as with control participants, results in the literature have commonly shown significantly greater moments among the most severe group (Sharma et al., 1999; Mundermann et al., 2005). Similarly, one prospective study showed that higher knee external adductor moments at baseline significantly predicted faster knee osteoarthritis progression (Miyazaki et al., 2002), and Astephen and colleagues showed a significant increase in the knee external adductor moment between a group of asymptomatic, moderate, and severe osteoarthritis patients (Astephen, Deluzio, Caldwell & Dunbar, 2008).

However, the osteoarthritis patients involved in these studies, even at baseline have frequently been categorised within the moderate to latter stages of disease progression, and in a systemic review Foroughi and colleagues highlighted the lack of research performed with early stage osteoarthritis participants (Foroughi et al., 2009). This has

lead to a theory suggesting that the increased knee external adductor moments found among osteoarthritis patients could be a consequence of the disease presenting at the knee joint (Foroughi et al., 2009). It has been previously suggested that higher knee external adductor moments among more severe cases of osteoarthritis could be a consequence of morphological changes such as cartilage loss and meniscus degeneration (Grainger & Cicuttini, 2004).

The results from this current investigation among the mature females, however tend to discount this possibility and maintain the initial theory suggesting the knee external adductor moment as a possible risk factor for osteoarthritis occurrence. The results from the examination of the knee joints indicated no loss of cartilage or joint space narrowing; factors which could have increased knee joint varus/valgus positioning (Grainger & Cicuttini, 2004). Similarly, all subjects were classified pain free, limiting the possibility of pain induced malalignment. The supposed theory suggesting knee external adductor moments as a possible consequence of knee osteoarthritis does not therefore seem applicable here. This is further supported by one prospective study that showed knee external adductor moment at baseline to be significantly correlated to initiation and progression of knee joint osteoarthritis (Miyazaki et al., 2002). Therefore, the results from the mature females alongside this prospective study tend to support the theory that the knee external adductor could be a risk factor for osteoarthritis development and progression.

6.4.5. Limitations

Within this magnetic resonance imaging study, several limitations were encountered. Firstly, the results of the magnetic resonance imaging scans, and the overall indications of osteoarthritis were based on a small sample size. Furthermore, of the ten participants that volunteered, one set of data had to be excluded due to metal occlusion in the knee joint. The difficulty in attaining volunteers was primarily due to a lack of desire to undertake a scan, and the incidence of claustrophobia among many of the mature females. This

limitation is not unusual and has been encountered in previous studies involving magnetic resonance imaging (Amin et al., 2005)

Secondly, the scans were only taken for the right knee of each participant, causing inferences to be made on the overall condition of each mature female. This was however based on the results from the biomechanical assessments, where data was only collected for the right knee of each participant. Collecting data for both knees would have doubled the scanning time, and could have lead to an additional drop in the number of volunteers.

Limitations were encountered in the positioning of the participant, and ability of each female to remain still for the duration of the scans. Occasionally, the scans appeared slightly blurred, indicative of movement within the magnet.

In addition, this study involved the use of a 1.5 T magnet, with T1 fat saturated images. However, fast spin echo T2 fat saturated images are more reliable for showing bone oedema, which could suggest that this lesion was present in the knees that were categorised as free from this feature (Pessis et al., 2003). Despite this, bone oedema was identified in three knees, and was therefore still included in the results.

6.5. Conclusion.

The overall findings from this magnetic resonance imaging study support the use of this technique to identify early stage osteoarthritis among asymptomatic mature female runners. Of the ten females that volunteered for participation, eight showed signs of early stage osteoarthritis, with the results from one participant inaccessible, and one female runner categorised as symptom free. These findings support the theories that osteoarthritis is prevalent among the more active mature members of the female population. Although non significant, a linear relationship between knee osteoarthritis and KOOS score indicates a tendency for mature females to recognise pain, however all females were categorised as symptom free. The significant relationship between knee osteoarthritis and knee external adductor moment strongly supports the theory that, in high magnitudes, this variable can lead to degradation of the knee joint. The diagnosis of each female within the early stage of osteoarthritis development further supports this theory, suggesting that the knee external adductor moment may be a contributing factor to the initial development of the debilitating condition. Furthermore, the association between this biomechanical variable and osteoarthritis shown in this current study could provide support for the use of the knee external adductor moment as a possible diagnostic tool for early signs of the condition in asymptomatic mature female runners.

Chapter 7. Summary and Conclusion.

Due to the number of people participating in running as a recreational or competitive activity, there has been a proliferation of research regarding the mechanisms and potential management techniques for running related injuries and conditions. The majority of these methods involve the adaptation of biomechanical variables during gait, using the influence of footwear control.

It has been suggested in the literature that certain aspects of running biomechanics are associated with an increased risk for the development of running related injuries and conditions, and this incidence is greater for women compared to men (Taunton et al., 2002). Furthermore, it has been speculated that as the age of the runner increases, so too does this risk (Verzijl et al., 2002). As such, this research project began by investigating lower limb biomechanics in young and mature females to determine the influence of age on running gait. The influence of a motion control shoe was then assessed in mature and young females to reveal whether changes in footwear could positively influence lower limb biomechanics. Alternative and combined footwear interventions were then assessed to identify appropriate footwear devices specific for mature female runners, with the potential to reduce the risk of overuse conditions and injuries. Finally, an MRI study occurred to investigate the current condition of the knee joint among a select group of mature female runners. This enabled further investigation of the association between biomechanical variables of gait and the development of knee joint osteoarthritis.

7.1. A Biomechanical Comparison of the Running Gait of Mature and Young Female Runners.

There has been a wealth of studies published to argue that biomechanical variables of running are linked with injuries and overuse conditions (Sharma et al., 2001; Taunton et al., 2002, Baliunas et al., 2002; Noehren et al., 2007). Specifically, rearfoot eversion and the coupled movement of knee internal rotation have received a vast level of attention, although the conclusions are still varied (Messier & Pittala, 1988; Hargrave. Carcia,

Gansneder, & Shultz, 2003). Additionally, considering osteoarthritis as an overuse condition (Van Gent et al., 2007) the knee external adductor moment has been consistently linked with its initiation, progression and development (Miyazaki et al., 2003).

Study One was performed to directly compare the running gait of a group of young (18-25 years) and mature (40-60 years) female runners. The experimental protocol was selected, on which the methodology of all further studies would be based. Since running is considered to be a bilaterally cyclic activity, for the purpose of this research it was speculated that there was a symmetry between left and right legs (Zifchock et al., 2008). Therefore both kinematic and kinetic data were collected for the right leg of each mature female performing running trials at a controlled speed. An assumption continued throughout each study of this research project was that rearfoot eversion can be determined from placing markers on the surface of the shoe as opposed to directly on the skin. It was therefore assumed that the foot does not move within the shoe, and that the rearfoot eversion angle calculated was not affected by the lacing or the fit of the heel around the calcaneum (Kilmartin et al., 1994).

Kinematic and kinetic analyses demonstrated that biomechanical measures could reveal differences in gait associated with age. Specifically, it was shown that during the stance phase of running, mature females display significantly greater magnitudes of peak loading rate, rearfoot eversion angle, knee internal rotation angle, and knee external adductor moment ($p<0.05$). Discussion and interpretation of these results suggested that these parameters of gait could be a contributing factor to the high levels of injuries and debilitating conditions among this specific group of mature female runners.

However a limitation of this investigation was highlighted in the footwear, with all females performing trials in their own footwear. As footwear variations have been shown to affect parameters of gait, it was deemed necessary to assess this difference in a controlled neutral shoe. A summary of conclusions based on the results form Study One are presented below:

- The running gait of mature and young females can be identified using three dimensional gait analysis.

- Significant differences exist in the biomechanics of running gait between young and mature females.
- Certain biomechanical features associated with mature female runners could predispose this group to running related injuries and conditions.

7.2. The effect of motion control shoes on the biomechanical running gait of mature and young female runners.

Study Two employed a similar methodology to that used in Study One, with a direct comparison between the running gait of mature and young females initially performed. However within this assessment, a controlled neutral shoe was worn by each participant to remove the possible effect of variations in footwear. In addition, a currently available motion control shoe was included to investigate the ability of footwear to control biomechanical variables of gait. Additional measures such as muscle strength and knee joint stiffness were also analysed to quantify and further explore gait and injury risk differences between the two groups. It was hypothesised that the neutral shoe would show differences in gait between the two groups similar to those found in Study One, and knee joint stiffness would be higher among the mature group. Secondly, it was hypothesised that the strength of the quadriceps femoris muscle would be lower among the mature group. Thirdly, it was hypothesised that the motion control shoe would reduce rearfoot eversion and knee internal rotation, and could therefore act as a method of injury management among the mature group. Finally, it was hypothesised that the motion control shoe would increase the knee external adductor moment during the stance phase of gait.

The neutral shoe was found to yield significantly greater rearfoot eversion, knee internal rotation, peak loading rates of impact force and knee external adductor moments among the mature group. The inclusion of a motion control shoe significantly reduced the rearfoot eversion and knee internal rotation angles among the mature group, although no difference was seen in the loading rate of impact force or the knee external adductor moment among either group. The assessment of muscle strength elicited no significant

difference between the two groups. Knee joint stiffness was however shown to be significantly greater among the mature group.

Discussion and interpretation of these findings lent support to Study One, substantiating previous suggestions that certain parameters of running gait change with age. Irrespective of footwear, mature female runners appear to adopt a running style that could predispose them to injuries and the development of debilitating conditions at the knee joint

The increase in knee joint stiffness displayed by the mature group lends support to the theory that this group is at an increased risk of developing debilitating condition at the knee joint (Griffin & Guilak, 2005). The ability of a motion control shoe to reduce rearfoot eversion and knee internal rotation supports previous research that concludes footwear modifications to play an important role in injury management (Rose et al., 2011). The significant reduction in rearfoot eversion and knee internal rotation among the mature group contests the theory that mature people are less responsive to footwear adaptations compared to younger individuals (Kakihana et al., 2007). Furthermore the biomechanical changes demonstrated with the use of a readily available running shoe could provide valuable insight into possible methods of reducing the injury risk associated with increasing age. These results support the use of gait analysis for footwear recommendations among the mature female runners, and lend support to the use of motion control shoes in reducing subtalar joint pronation.

Given that the knee external adductor moment was not affected by the use of a motion control shoe, yet this variable is commonly associated with the development of debilitating conditions such as osteoarthritis, further investigation was required to determine whether this variable can be reduced with variations in support underfoot.

Summary conclusions:

- In a controlled neutral shoe, mature females display significantly different biomechanics of running gait compared to a younger group.
- The use of a currently available motion control shoe significantly reduced rearfoot eversion and knee internal rotation among the mature female runners.
- The motion control shoe did not affect the knee external adductor moment among either group of runners.

- Current motion control shoes are not sufficient to control for all aspects of gait specific to mature female runners that could predispose to overuse conditions.

7.3. The effect of medial and lateral wedges and an orthotic intervention on the running gait of mature female runners.

Study Three was developed based on the results from the previous two investigations. It had consistently been shown that mature females illustrated variables of gait that could predispose to injuries and conditions. Although the use of a current motion control shoe was shown to reduce peak rearfoot eversion and knee internal rotation compared to a neutral shoe in both young and mature females, the knee external adductor moment was considered to remain a high risk factor for the development of medial compartment knee joint osteoarthritis. In addition, the knee external adductor moment has also been linked with the development of other conditions, such as injury to the meniscus of the knee joint (Davies-Tuck et al., 2008), which in turn increases the risk of developing medial knee joint osteoarthritis in the future. This therefore substantiates the need to identify methods of reducing the knee external adductor moment.

The use of a medial wedge was hypothesised to produce changes similar to those shown by the motion control shoe in Study Two. In contrast the inclusion of a lateral wedge was expected to reduce the knee external adductor moment, due to the previously shown association between lateral wedge and reductions in this moment (Crenshaw et al., 2000). The additional intervention orthotic was included based on the results of Study One and Study Two, including both a medial and lateral wedge. It was therefore hypothesised that the medial wedge would produce reductions in the rearfoot eversion and knee internal rotation angles, and the lateral wedge would reduce the knee external adductor moment. It was hypothesised that the orthotic intervention would significantly reduce all three variable of gait.

Results and discussion showed little effect of the 6 mm lateral and medial wedges on the kinematics of running gait among mature female runners. As speculated within the discussion of Study 2, although not extensively reported within the literature, it is

possible that the level of support in the current running shoes worn regularly by each individual, and differences in individual biomechanical characteristics may have influenced the biomechanical responsiveness to different footwear interventions. However the intervention orthotic was shown to significantly reduce the rearfoot eversion, knee internal rotation and the knee external adductor moment during the stance phase of gait. This suggests that a combined effect of lateral wedging with medial arch support could be an appropriate intervention to reduce the risk of debilitating conditions among mature female runners.

Summary conclusions:

- A 6mm medial wedge based in the heel of the shoe did not significantly alter the biomechanics of running gait among a group of mature female runners.
- Although no change in kinematics was found with the 6mm lateral wedge, this intervention did significantly reduce the knee external adductor moment among mature female runners.
- An orthotic intervention incorporating both a medial and lateral wedge significantly reduced rearfoot eversion, knee internal rotation, and the knee external adductor moment among mature female runners.
- A combined intervention including both medial and lateral wedge technology may be appropriate to reduce injuries and overuse conditions among mature female runners.

7.4. The use of MRI scanning to identify osteoarthritis of the knee joint among mature female runners and associations with the knee external adductor moment.

The fourth study of this research thesis was included to assess the presence of osteoarthritis among a selection of mature female runners. It has been postulated that a poor link exists between the development of knee joint osteoarthritis and the development of symptoms (Ehrman et al., 2003). As such, the knees of a group of mature female runners were scanned using the magnetic resonance imaging techniques.

Results illustrated 8 from the 9 mature female runners to exhibit a minimum of one sign of early stage osteoarthritis, based on the presence of physiological changes within the joint. This therefore supports the use of this method in identifying early stages of the condition among asymptomatic individuals. Correlation tests between these results and the knee external adductor moment lent support to the association between high moments and osteoarthritis development among mature female runners. As such it was suggested that the biomechanics adopted by a group of mature female runners could lead to the initiation of debilitating conditions at the knee joint, even prior to the onset of symptoms. These results therefore lend support to earlier suggestions that footwear interventions are recommended as a method for injury management among the mature members of the female running population. Similarly, due to the strong association between the knee external adductor moment and the presence of early stage osteoarthritis at the knee joint, it is proposed that assessment of this variable using three dimensional gait analysis techniques could be implemented as a tool in early detection of the degenerative condition.

Summary conclusions:

- Although the KOOS survey showed all females to be free from symptoms of osteoarthritis, the results from an MRI scan showed 8 out of 9 mature female runners to present signs of early stage osteoarthritis.
- The significant correlation between the knee external adductor moment and the identified signs of knee joint osteoarthritis supports this variable as an indication of joint loading, and a possible tool in diagnosis of early stages of the condition.

7.5. Summary and future direction.

The main purpose of this research project was to investigate biomechanical changes in running style among the mature members of the female population. Due to the lack of research regarding this specific group of runners, the results from this project increase the area of knowledge regarding the link between age and injury or osteoarthritis risk. Understanding the role of footwear in injury management methods is vital, due to the effect on the kinematics and kinetics during motion and the accessibility of this intervention. The results from this project lend support to the theory that footwear interventions can alter biomechanical variables of running gait among a mature group. Furthermore, the promising results from the third study suggest that an intervention orthotic incorporating both medial arch support and a lateral wedge is appropriate to limit the risk of injuries and conditions associated with the gait of mature female runners. Although the results do not directly support footwear as an injury prevention method, the abundance of literature that has associated certain biomechanical features of gait with injuries lends support to the theory.

Future research within this area is required to investigate the possible causative factors leading to the changes in running biomechanics among mature females. Although the physiological variable of muscle strength was assessed, and the concept of hormonal fluctuations was speculated upon, an increased depth of research is required. Furthermore, within the specific field of osteoarthritis, future research is required among mature female runners to compare between those free from osteoarthritis and those diagnosed with early or variations in the stage of the condition, to further assess the effect on the biomechanical variables of running gait. Longitudinal research is also required, to assess the strength of relationship between the identified variables of gait, and the development of injuries and conditions over time.

In terms of footwear development, results from the mature females elicited promising inferences. Although motion control shoes currently exist to reduce subtalar joint pronation among runners, this intervention does not appear sufficient for the mature female runners, to control for the knee external adductor moment. Within the running industry, at present the categories of footwear include the groups; males, females, and children. Conclusions drawn from this study suggest that the current range of footwear

may not accommodate for all biomechanical changes in running gait that occur with increasing age among the female population. Furthermore, among the mature females in the current study, biomechanical movement patterns have been shown to change throughout the stance phase of gait, suggesting variable support is required in different areas of the shoe sole. As a cohort of runners comprising high numbers, it is suggested that mature female runners would benefit from specific footwear interventions that would reduce the likelihood of developing overuse injuries and conditions. It is proposed that footwear companies give consideration to the development of an additional category of running shoe.

APPENDIX

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Appendix A: Peak Vicon Accuracy Study.

Introduction and Literature Review.

Introduction.

Within the domain of scientific analysis, the quality of the data achieved is measured against a number of factors; accuracy, relevance, timeliness, completeness, trust and accessibility (Olsen, 2003). Data accuracy is a measurement of error, and is therefore the dimension that formulates the foundation of data quality. In all applications of scientific measurement, the reproducibility and accuracy of the method must be established in order to enable meaningful contribution (Eston & Reilly, 2001). Assessing and making large improvements in data accuracy and the quality of information from the data can however only be accomplished through proactive activities. Therefore, within biomechanical research, accuracy assessments are essential, as without highly accurate data, information quality cannot be achieved (Olsen, 2003).

Movement Analysis.

The most commonly used gait tool in the clinical setting is observational gait analysis (Montgomery & Connolly, 2002). This application requires systemic observation of each body segment throughout each phase of gait, and has a number of limitations (Chapter 2). When in combination with one or more force platforms, three dimensional motion analysis systems provide the main implement in gait analysis (Lewis, Stewart, Postans & Trevelyan, 2007). Analysis of gait was traditionally used as a research tool; however the impetus to diagnose and objectively document the benefits of medical intervention has resulted in a reliance on such analysis as a clinical diagnostic tool (Skinner, 1995). The assessment and publication of normal data allows for definition of pathological gait, and analysis in a gait laboratory can assess initial deviation from “normal” values. Therefore,

as an increasingly common procedure in the diagnosis of movement disorders, and as a major determinant of important clinical decisions, confidence in the quality of gait assessment is essential. These systems therefore require regular monitoring to detect changes in performance; a possible consequence of sudden changes such as camera failure or slow deterioration (Lewis *et al*., 2007).

The worthwhile application of data collected through gait analysis is therefore dependent on the resulting accuracy and precision of the testing system (Payton & Bartlett, 2008). During motion analysis, measured values invariably differ from the actual values; they contain errors. Therefore, quantification of the errors in a measurement is essential, to ensure the certainty with which a statement about the results of an analysis is known (Payton & Bartlett, 2008).

Calibration.

The process of calibration involves comparing an instrument's performance to a standard of known accuracy; to assess and document any deviation from this known standard. For any system collecting kinematic data, a suitable means of calibration must be used to ensure the image co-ordinates are correctly scaled to size (Robertson, Caldwell, Hamill, Karren, & Whittlesey, 2004). Although errors cannot completely be eliminated, regular calibration checks are essential to minimise the potential errors in a measurement; forming part of a quality assurance programme (Lewis, *et al*., 2007). Without calibration, it is impossible to measure distances accurately (Whittle, 2003). The calibration process exploits a model or calibration frame which is viewed by all the cameras. Computer software is then used to calculate the relationship between the known three dimensional position of the markers on the frame, and the two dimensional positions of those markers in the fields of view of the different cameras (Whittle, 2003). For two dimensional data analysis, a calibrated ruler is placed in the plane of motion, and its known length digitised. However, within three dimensional data multiple cameras are employed, and a set of "control points" are established whose exact coordinates are known (Robertson *et al*., 2004). As suggested, calibration does not eliminate, but minimises error in

measurement, and within calibration of a motion analysis system, the control point positions are not known to 100% accuracy (Payton & Bartlett, 2008).

Error.

Performance of motion capturing systems strongly depends on their setup and is highly sensitive against alterations (Windolf, Gotzen, & Morlock, 2008). During data collection, performance errors arise through various situations including marker propertied and camera configuration (Furnee, 1991). Analysis of error is fundamental to the design of any instrument; an understanding of the source of error enables insight into the nature of the scientific measurement under analysis (Kamm, 1995). Both two-dimensional as well as three-dimensional analyses are subject to error (Ackland, Elliott, & Bloomfield, 2008). Initially, research into the errors associated with the three-dimensional system has primarily focussed on the comparison between opto-reflective and video-based systems (Ackland *et al*., 2008). However, it is well known that several error sources exist associated with both kinematic and kinetic data analysis, often attributed to limitations of the hardware and software capabilities (Dorocjak & Cuddeford, 1995). In general, error is introduced due to the process used to collect data (Grimshaw, Less, Fowler, & Burden, 2007). System performance errors include camera resolution and placement which affect the quality of the image.

An additional common source of error is that due to a change in condition at the time of acquisition compared with when calibration occurred (Whittle, 2003). Finally, other sources of error can include skin movement, mathematical model assumptions and marker placement (Dorocjak & Cuddeford, 1995). According to Kamm *et al*., (2007), errors during data collection for motion analysis can be categorised into image recording errors, digitising errors and timing errors. Image recording errors include foreshortening error, depth error and obliquity error, suggesting the image is portrayed as smaller, greater or more blurred at the edges than in its true form. Digitising error refers to point location error, where difficulty exists in identification of reference points and joint centres. Additionally, resolution errors occur, due to the ability of the digitising system to resolve,

the image size, and the size of the field of view. Finally, timing errors generally arise due to the timing mechanism used, and the timing error, for example, heel strike in running can only be judged to ±1 frame (Kamm *et al*., 2007). The overall effect of the above errors is a general inexactness with which any single point can then be estimated. Quantification of this error is therefore required via an assessment of the accuracy of the system. The main aim of this study is therefore to assess the level of accuracy of the Peak Vicon Optical Capture System in the University of Exeter Biomechanics Research Laboratory.

Methods.

Introduction to methods.

Accuracy is assessed through a comparison between known and measured values; during image-based motion analysis, accuracy is reported as the difference between the true location of the control points and their predicted values (Payton & Bartlett, 2008). The control points utilised in this assessment are however different from those used during the calibration process. According to Challis and Kerwin (1992), to obtain a true estimation of accuracy, an independent assessment criterion is required as utilisation of the same points greatly overestimates accuracy. Therefore a frame or accuracy model is utilised specific for the accuracy assessment. The performance of three dimensional measurement tools is quantified with both static and dynamic tests (Scheirman and Aoki 1999), the former assesses accuracy from stationary points within a specified space, and the latter from moving points. Within this study, both methods will be employed and results analysed separately.

Calibration.

In advance of the accuracy assessment procedure, the eight cameras employed in the motion analysis system were calibrated. Four retroreflective markers, arranged in an L shape on a frame placed on the floor, were used for setting up the Vicon coordinate system at static calibration (Windolf *et al*., 2008). The frame covered an operation volume of 1.0 by 0.5 by 0.75 m^3. It was essential that all markers were in view of each camera to be calibrated. Following this setup phase, a calibration wand was used, with two fixed markers placed 0.913m apart. This wand was then oscillated in the field of view of all cameras, within the area outlined by the static frame. Data was collected for 120 seconds, with the wand moving throughout the entire cubic area. Once the system was

successfully calibrated, the newly calibrated template was imported into a template prepared for the accuracy assessment procedure.

Static Assessment.

The static accuracy assessment took place in the biomechanics research laboratory at the University of Exeter. This assessment involved the use of a modified multi-arm frame (Figure A.1) situated in the field of view of all 8 cameras. As illustrated in Figure A.1, this frame included 16 reflective markers or spheres of known coordinates, located along 8 rods. The cameras were kept stationary while the multi-arm frame was videoed. The calibration frame with four retro reflective markers arranged in an L shape, was once again used, increasing the ease of marker identification during the data processing stage.

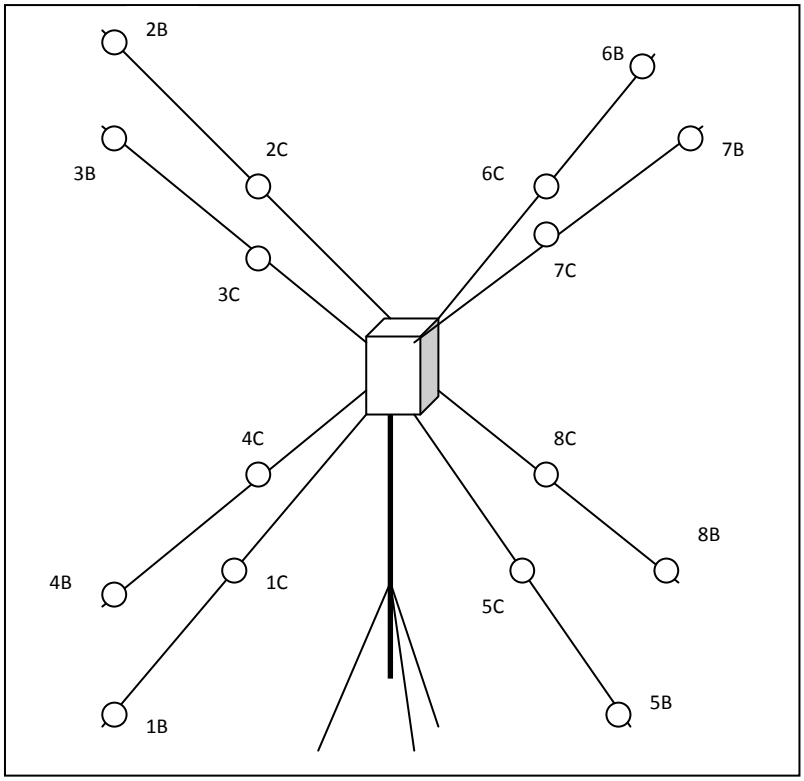


Figure A.1. Illustration of the Multi-Arm Frame used in the static Accuracy Assessment.

The view from each camera was checked, ensuring the frame was visible. The presence of additional unknown markers was assessed; any apparent were minimised through reduction of light and eclipsing reflective areas. Following the setup phase, with all cameras turned on and all markers in the field of view of each, the data collection phase was instigated. The recording lasted a total of 5 seconds, with no movement in the field of view. The Peak Vicon software was used to transfer images recorded from each camera, and present raw data on the computer. The template used in the Vicon system was specifically designed for the static accuracy assessment, therefore enabling each marker to be assigned its correct label. The paths were then identified manually, ensuring each marker was represented with its specific label and subsequent coordinates.

Transformed co-ordinates were then assessed. The coordinates of each marker were calculated relative to the origin (0,0). Data were then compared to a table of accurate coordinates for each marker in the x, y, and z directions.

Dynamic Assessment.

Similar to the static assessment, the dynamic assessment of the accuracy of the Vicon motion analysis system took place in the research laboratory at the University of Exeter. In contrast to the static assessment, a wand was moved within the field of view during data collection. This procedure involved the use of a 1 meter long black aluminium bar with matte finish. A marker was attached at each end of the 0.913m calibration wand. The exact distance was measured prior to the measurement, to ensure no unwanted marker movement had occurred. Similar to the static assessment, the L shaped calibration frame was positioned on the floor to increase the ease of marker identification and direction of camera view. Once again, a template specifically designed for the dynamic accuracy assessment was opened, and the newly calibrated template was imported. The view from each camera was checked to ensure the calibration frame markers were clearly visible, and any unwanted reflections were minimised. The dynamic accuracy assessment then occurred in the three planes of motion; Vertical (z), Horizontal (y) and Transverse (x).

The measurement of accuracy in the vertical direction was the first determined. The aluminium bar was held in the subject's hand, ensuring their forearm was parallel with their torso. A cardanic joint was attached to the centre of the wand, allowing free oscillation during data collection. Using this joint, the wand was then moved in the vertical direction, with minimal deviations in either the x or y planes of motion. The wand was moved an approximate distance of 0.3meters, with data collected for a total of 5 seconds. This test was then performed four more times, collecting a total of five separate sets of data to improve reliability.

The measurement of accuracy in the direction of motion across the force plate (y) was then determined. The subject walked along the field of view of the cameras, holding the wand by the joint, in the same position as was described for the vertical test. The subject walked forwards and backwards at an extremely slow pace, to ensure the wand was in the field of view of the cameras for all 5 seconds of data collection. Once again, this test was then performed four more times, producing 5 sets of data.

Finally, measurement of accuracy in the medio-lateral direction (x) direction was determined. The subject once again positioned the wand, their forearm parallel to their torso, and walked across the plate in the x direction with the wand held out in front. Again, movement in the z and y directions was kept as minimal as possible. Data was collected for 5 seconds, with all markers in the field of view for the entire duration. As occurred for the two previous dynamic accuracy tests, a total of 5 different tests occurred, producing 5 consecutive sets of data. All data from each test performed in this accuracy assessment was then transferred into an excel spreadsheet, and calculations to determine the distance between the markers were performed.

Root Mean Square Error.

Following completion of the data collection stage, a series of mathematical and statistical calculations were completed to process and apply the results. A main method for assessing the accuracy of a system is through the arithmetic error; the mean difference between the measured and the true values. Although this method does provide information regarding the system accuracy, it is not without fault, as both negative and positive values within the data can cancel one another out (Payton & Bartlett, 2008). The preferable measure of accuracy is the root mean square error as it is the most conservative of the criterion (Payton & Bartlett, 2008). This measurement statistic for accuracy was similarly used in an early research study by Challis and Kerwin (1992), and is calculated through taking many measurements of a point location (Scheirman & Aoki, date).

$$\sqrt{\sum(X-\alpha)^2/(n-1)}$$

Equation A1.

Where:

X	=	Measured Value
α	=	True Value
n	=	Number of times measurement was taken

The above algebraic equation describes the process for determining the root mean square error of a data set. The value achieved describes the overall residual error between data points, highlighting the accuracy of the measurement system (Mow & Huiskes, 2004).

Results.

Static Accuracy Assessment.

Within the static assessment of accuracy, a series of five consecutive data sets were produced from five individual tests. These data were then transferred into an excel spreadsheet and analysed. Three coordinates (x, y, and z) were produced for each marker; each measured relative to the assigned origin (0,0). The data were then compared with known true values. Two tables of results are presented, indicating the measured values and the standard expected coordinate values (Table A.1), and the calculated root mean square errors (Table A.2).

Table A1. Summary table of results from the static accuracy assessment, illustrating calculated and known coordinates of each marker.

ROD		X COORDINATE (mm)		Y COORDINATE (mm)		Z COORDINATE (mm)	
NUMBER	MARKER	Measured Value	Real Value	Measured Value	Real Value	Measured Value	Real Value
1	B	0.349	0.3446	0.4817	0.4813	0.4223	0.4224
	C	0.594	0.5920	0.8262	0.8268	0.7232	0.7232
2	B	0.3408	0.3408	0.4784	0.4783	1.504	1.5042
	C	0.5913	0.5913	0.8264	0.8265	1.22	1.2100
3	B	0.3396	0.3395	1.7445	1.7446	1.54002	1.54002
	C	0.5908	0.5908	1.3949	1.3949	1.208	1.2080
4	B	0.345	0.3449	1.737	1.7370	0.4212	0.4211
	C	0.5915	0.5918	1.3934	1.3934	0.722	0.7220
5	B	1.2415	1.2414	1.7393	1.7393	0.4202	0.4201
	C	0.996	0.9962	1.3934	1.3934	0.7215	0.7214
6	B	1.2407	1.2484	1.75	1.7438	1.5022	1.5022
	C	0.9976	0.9988	1.393	1.3950	1.209	1.2090
7	B	1.2428	1.2428	0.4771	04770	1.505	1.5044
	C	0.9988	0.9988	0.8253	0.8253	1.2096	1.2095
8	B	1.2427	1.2428	0.479	0.4791	0.4223	0.4223
	C	0.9968	0.9967	0.9967	0.9967	0.7232	0.7232

Table A2. Summary table of the root mean square errors across all markers along each axis.

Root Mean Square Error (RMSE) (m)		
X	Y	Z
0.002	0.001	0.002

Dynamic Accuracy Assessment.

Within the dynamic assessment of accuracy, five consecutive trials were achieved in each direction of motion (x, y, and z). The average value produced from a total of 5 seconds of data collection was calculated and compared to the real value (0.913). These data are summarised in Table A.3, illustrating average results. Root Mean Square Error calculations then occurred to determine how close the data produced was to the true values (Table A.4). A demonstration of this root mean square error result is demonstrated in the graph, illustrating the fit between the calculated data line and the standard. Figure A.2 presents a sample data set in graphical form. The line graph shows data collected in the vertical (z) direction on trial 3. The calculated wand length is plotted against the standard length, illustrating minimal deviations.

Table A3. Summary table of results from the dynamic accuracy assessment. Values report distance measured between coordinates of two markers.

TRIAL NUMBER	REAL DISTANCE (m)	WAND MOVEMENT IN DIRECTION		
		X	Y	Z
1	**0.913**	0.913	0.915	0.914
2		0.91	0.912	0.910
3		0.910	0.911	0.912
4		0.912	0.913	0.91
5		0.915	0.910	0.912
AVERAGE		**0.912**	**0.912**	**0.911**

Table A4. Summary table of the root mean square errors (m).

TRIAL	X	RMSE	Y	RMSE	Z	RMSE
1	0.913	<0.0005	0.915	0.009	0.914	0.001
2	0.91	0.003	0.912	0.002	0.910	0.003
3	0.910	0.003	0.911	0.003	0.912	0.002
4	0.912	0.002	0.913	<0.0005	0.91	0.003
5	0.915	0.009	0.910	0.003	0.912	0.002

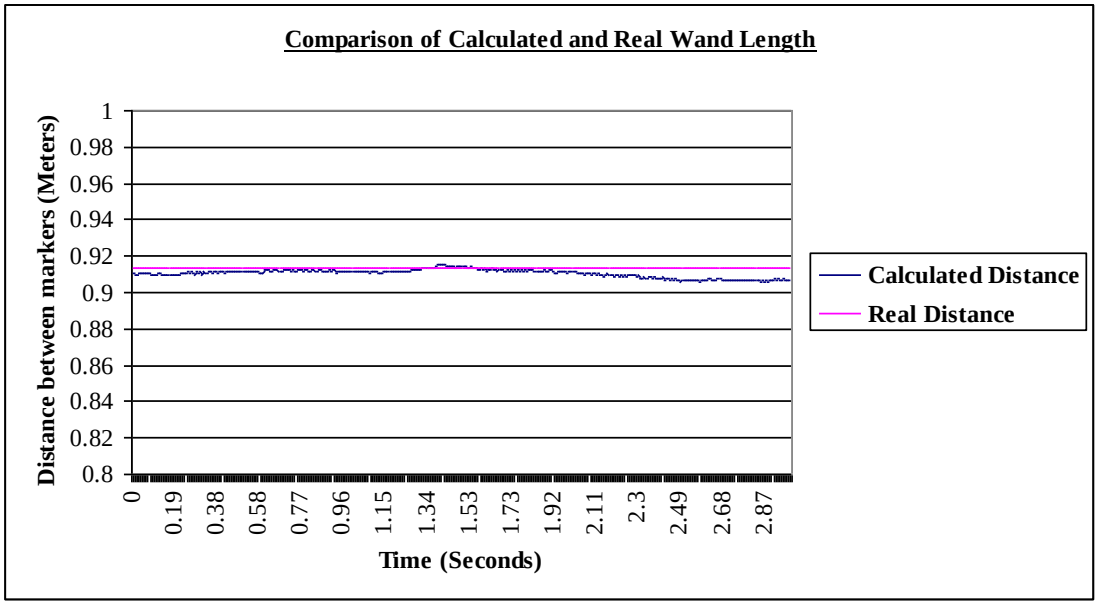


Figure A.2. Sample line graph illustrating comparison between calculated and real wand length. Data taken from trial 3 in the dynamic accuracy assessment with wand movement in the Z direction.

Discussion and Conclusion.

Static Accuracy Assessment.

The principle aim of the current study was to assess the accuracy of the Peak Vicon Optical Capture System within the University of Exeter Biomechanics Research Laboratory. Two separate tests were performed; a static and a dynamic assessment, and the errors associated with both were investigated. Within the static assessment of accuracy, three-dimensional coordinates of 16 stationary reflective markers were calculated. A modified multi-arm frame was employed, as illustrated in the methods section, and the locations of markers along each rod (arm) were calculated in relation to the origin sphere. Mean difference of the measured coordinates compared with the true values were computed and the root mean square errors (RMSE) were assessed. This statistic measures both the deviation and consistency of results, and provides a clear indication of the accuracy of the measurements. Greater values are associated with higher deviations, relating to decreased accuracy (Schmidt & Lee, 1999). In this study, root mean square error of 0.002, 0.001 and 0.002 in the X, Y and Z directions respectively were found for reconstruction accuracy of fixed test points in the calibrated volume. These results compare well to those presented by Scheirman and Aoki (1999), who similarly assessed the accuracy of the Peak Optical Motion Capture System. The RMSEs presented in their results were reported as 0.001, 0.002 and <0.0005 in the X, Y and Z directions respectively. Additionally, when assessing the accuracy of the larger (in comparison to the current study) 6.6m^3 volume, Coleman and Rankin (2005) illustrated root mean square errors of 0.007, 0.009 and 0.005 in the X, Y and Z directions respectively following a static assessment with a calibration frame. Finally, the results from the present study suggest a high level of accuracy was achieved, when compared to data from the research of Challis (1995). This study reported root mean square errors ranging from 0.006 to 0.023 in the similar coordinates measured; higher values than those reported in the present study.

Subsequently, the results from the static accuracy assessment performed in the present study combined with the weight of the research, suggests the peak vicon optical capture

system produced data with a high level of accuracy. With a maximum difference of only 3mm, compared with 20mm previously reported by other researchers, the system is suggested accurate for the purpose of the data collection methods.

Dynamic Accuracy Assessment.

The data achieved from the dynamic accuracy assessment produced evidence of system performance and identified possible failures (Lewis *et al*., (2007). This notion is supported by Sheirman and Aoki, (1999) who stated dynamic accuracy assessments to reveal errors occurring from movements. It is therefore essential to determine the inherent error and overall accuracy of any experimental procedure (Anderson, 2009). The dynamic accuracy analysis in this study was carried out within an observation volume as typically used for biomechanical applications investigating motion magnitudes (Windolf, *et al*., 2008). This test was employed to assess the accuracy of the Peak Vicon Motion Analysis system involved the use of an aluminium wand, affixed with a reflective marker at each end. This wand was then moved in the three planes of motion (x, y and z), producing three separate sets of data. Within each data set, the coordinates of the markers were identified, and the measured distance between the two calculated. This figure was then compared to the standard known length of the wand (0.913m), and the presence of differences assessed. The calculated magnitude of the differences between figures then produced an indication of the accuracy of the system.

The root mean square errors of the results from each trial are expressed in Table A.4 in the results section. The mean RMSE for each coordinate, X, Y and Z are illustrated, suggesting an estimation of the overall accuracy of the system for moving points. As illustrated, the average RMSE range from 0.002 to 0.003, with mean errors of 3mm, 3mm and 2mm in the X, Y and Z direction. These results are similar to those suggested in the literature regarding the static accuracy assessments, and were once again lower than the results of the study performed by Coleman and Rankin, (2005), suggesting a higher level of accuracy. Furthermore, when compared with results of an earlier study performed by Ehara, Fujimoto, Miyazaki, Mochimaru, Tanaka, and Yamomoto (1997), the results from

the present study once again indicate a high level of accuracy. Within their investigation, Ehara *et al*., (1997) demonstrated errors ranging from 4mm to 8.5mm, when dynamically assessing the Vicon Optical Motion Capturing System in the three coordinates of movement, X, Y and Z. Therefore, with one maximum error value of 9mm, and an average of 3mm, the values in the present study suggest an acceptable level of accuracy achieved.

Conclusion.

Three dimensional motion analysis systems are widely used clinical tools. When in combination with one or more force platforms, these systems provide the main implement in gait analysis. This study was conducted to examine the accuracy of observational methods for analysis of the lower limb during running gait. The data presented above suggests that the Peak Vicon Optical Motion Capturing System worked effectively, producing accurate data with minimal error. As suggested, no motion capturing system can work perfectly; a degree of error will always be present. The presence of error causes the measured values to invariably differ from the actual true values; however it is the magnitude of the errors that determines the accuracy of the system. Although the results combined with the weight of the research suggest the system under investigation produced effectively accurate data, errors still exist.

Attributes such as marker roundness or disfigurement, reflective capacity, optical contrast, lighting conditions, calibration velocity and duration, and size of the field of view may all affect measuring error (Windolf *et al*., 2008). Within their assessment of accuracy, Windolf *et al*., (2008) revealed larger markers to significantly increase the level of accuracy and precision of the motion capturing system, with each marker size establishing a statistical subset. Additionally, within the dynamic assessment, although kept as constant as possible, manual wand motion remains uncontrolled. The use of an automated calibration procedure may therefore increase the accuracy due to a more consistent calibration path (Windolf *et al*., 2008).

In order to assess the effect of such error present in the motion analysis system, future research should involve the performance of a sensitivity study. This will determine what effect the maximum possible error will have on data achieved. However, through comparisons with previous studies, it is suggested that outcome data from this system is accurate for the purpose of biomechanical evaluation of lower limb motion during running.

Appendix B

Knee and Osteoarthritis Outcome Score (KOOS), English version LK1.0 1

KOOS KNEE SURVEY

Todays date: _____/_____/_____ Date of birth: _____/_____/_____

Name: _____

INSTRUCTIONS: This survey asks for your view about your knee. This information will help us keep track of how you feel about your knee and how well you are able to do your usual activities.
Answer every question by ticking the appropriate box, only one box for each question. If you are unsure about how to answer a question, please give the best answer you can.

Symptoms
These questions should be answered thinking of your knee symptoms during the **last week**.

S1. Do you have swelling in your knee?

Never Rarely Sometimes Often Always
o o o o o

S2. Do you feel grinding, hear clicking or any other type of noise when your knee
moves?

Never Rarely Sometimes Often Always
o o o o o

S3. Does your knee catch or hang up when moving?

Never Rarely Sometimes Often Always
o o o o o

S4. Can you straighten your knee fully?

Always Often Sometimes Rarely Never
o o o o o

S5. Can you bend your knee fully?

Always Often Sometimes Rarely Never
o o o o o

Stiffness
The following questions concern the amount of joint stiffness you have experienced during the **last week** in your knee. Stiffness is a sensation of restriction or slowness in the ease with which you move your knee joint.

S6. How severe is your knee joint stiffness after first wakening in the morning?

None Mild Moderate Severe Extreme
o o o o o

S7. How severe is your knee stiffness after sitting, lying or resting **later in the day**?

None Mild Moderate Severe Extreme
o o o o o

Pain

P1. How often do you experience knee pain?

Never Monthly Weekly Daily Always
o o o o o

What amount of knee pain have you experienced the **last week** during the following ativities?

P2. Twisting/pivoting on your knee

None Mild Moderate Severe Extreme
o o o o o

P3. Straightening knee fully

None Mild Moderate Severe Extreme
o o o o o

P4. Bending knee fully

None Mild Moderate Severe Extreme
o o o o o

P5. Walking on flat surface

None Mild Moderate Severe Extreme
o o o o o

P6. Going up or down stairs

None Mild Moderate Severe Extreme
o o o o o

P7. At night while in bed

None Mild Moderate Severe Extreme
o o o o o

P8. Sitting or lying

None Mild Moderate Severe Extreme
o o o o o

P9. Standing upright

None Mild Moderate Severe Extreme
o o o o o

Function, daily living

The following questions concern your physical function. By this we mean your ability to move around and to look after yourself. For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A1. Descending stairs

None Mild Moderate Severe Extreme
o o o o o

A2. Ascending stairs

None Mild Moderate Severe Extreme
o o o o o

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A3. Rising from sitting

None Mild Moderate Severe Extreme
o o o o o

A4. Standing

None Mild Moderate Severe Extreme
o o o o o

A5. Bending to floor/pick up an object

None Mild Moderate Severe Extreme
o o o o o

A6. Walking on flat surface

None Mild Moderate Severe Extreme
o o o o o

A7. Getting in/out of car

None Mild Moderate Severe Extreme
o o o o o

A8. Going shopping

None Mild Moderate Severe Extreme
o o o o o

A9. Putting on socks/stockings

None Mild Moderate Severe Extreme
o o o o o

A10. Rising from bed

None Mild Moderate Severe Extreme
o o o o o

A11. Taking off socks/stockings

None Mild Moderate Severe Extreme
o o o o o

A12. Lying in bed (turning over, maintaining knee position)

None Mild Moderate Severe Extreme
o o o o o

A13. Getting in/out of bath

None Mild Moderate Severe Extreme
o o o o o

A14. Sitting

None Mild Moderate Severe Extreme
o o o o o

A15. Getting on/off toilet

None Mild Moderate Severe Extreme
o o o o o

For each of the following activities please indicate the degree of difficulty you have experienced in the **last week** due to your knee.

A16. Heavy domestic duties (moving heavy boxes, scrubbing floors, etc)

None Mild Moderate Severe Extreme
o o o o o

A17. Light domestic duties (cooking, dusting, etc)

None Mild Moderate Severe Extreme
o o o o o

Function, sports and recreational activities

The following questions concern your physical function when being active on a higher level. The questions should be answered thinking of what degree of difficulty you have experienced during the **last week** due to your knee.

SP1. Squatting

None Mild Moderate Severe Extreme
o o o o o

SP2. Running

None Mild Moderate Severe Extreme
o o o o o

SP3. Jumping

None Mild Moderate Severe Extreme
o o o o o

SP4. Twisting/pivoting on your injured knee

None Mild Moderate Severe Extreme
o o o o o

SP5. Kneeling

None Mild Moderate Severe Extreme
o o o o o

Quality of Life

Q1. How often are you aware of your knee problem?

Never Monthly Weekly Daily Constantly
o o o o o

Q2. Have you modified your life style to avoid potentially damaging activities to your knee?

Not at all Mildly Moderatly Severely Totally
o o o o o

Q3. How much are you troubled with lack of confidence in your knee?

Not at all Mildly Moderately Severely Extremely
o o o o o

Q4. In general, how much difficulty do you have with your knee?

None Mild Moderate Severe Extreme
o o o o o

Thank you very much for completing all the questions in this questionnaire.

Appendix C: Study One Individual Results.

C.1. Individual Q angle data for mature and young female runners (Study 1).

Participant	Q Angle	
	Young	Mature
1	5.477	1.386
2	6.74	2.506
3	0.346	25.873
4	2.997	19.42
5	26.674	9.41
6	1.48	10.48
7	9.675	4.64
8	3.75	2.912
9	0.36	4.096
10	8.64	13.65
11	5.635	11.46
12	6.94	8.736
13	9.75	2.313
14	5.75	3.476
15	5.08	0.147
MEAN (Standard Deviation)	**6.62 (±6.3)**	**8.03 (±7.29)**
P-Value	**P = 0.57**	

C.2. Individual mean data for mature and young female runners (Study 1).

Pk Loading Rate (BW/s)		Pk RF Eversion (deg)		Pk Knee Int Rot (deg)		Peak DF (deg)		Pk KEAM (deg)	
M	Y	M	Y	M	Y	M	Y	M	Y
50.6 ± 5.6	36.24 ± 8.8	5.4 ± 3.2	5.21 ± 1.6	9.3 ± 4.2	13.9 ± 6.2	21.9 ± 8.1	22 ± 2.6	2.079 ± 0.39	1.1 ± 0.23
75.1 ± 10.2	47.104 ± 10.7	3.6 ±1.96	6.8 ± 1.3	9.5 ± 3.9	10.02 ± 8.5	24.8 ± 4.8	30.3 ± 7.72	2.08 ± 0.1	1.3 ± 0.15
59.8 ± 9.6	39.304 ± 6.7	8.6 ± 2.4	9.3 ± 3.46	10 ± 3.9	4.43 ± 2.8	28.2 ± 2.9	30.5 ± 8.1	1.97 ± 0.44	1.48 ± 0.16
59.86 ± 7.1	17.4 ± 7.8	10.2 ± 5.2	4.28 ± 3.2	12 ± 4.5	5.77 ± 4.2	29.5 ± 3.2	28.7 ± 2.0	2.01 ± 0.53	1.18 ± 0.37
51.24 ± 10.0	51.04 ± 9.9	12.5 ± 1.6	3.29 ± 4.6	16 ± 2.5	5.64 ± 2.2	19.99 ± 3.21	28 ± 7.6	2 ± 0.19	1.66 ± 0.15
54.16 ± 6.3	44.104 ± 3.5	12.5 ± 2.33	3.21 ± 5.4	10.8 ± 2.66	4.74 ± 1.8	22.5 ± 1.25	29.2 ± 5.31	1.2 ± 0.22	0.96 ± 0.5
64.51 ± 9.8	36.48 ± 4.8	11.2 ± 3.49	11.4 ± 5.6	14 ± 2.4	9.12 ± 5.7	21.7 ± 2.8	30.25 ± 4.9	1.46 ± 0.3	1.96 ± 0.49
56.92 ± 8.5	36.48 ± 5.3	10 ± 4.2	5.67 ± 0.9	11.4 ± 1.96	10.3 ± 2.9	21.0 ± 4.6	28 ± 5.2	1.64 ± 0.8	1.16 ± 0.30
46.87 ± 8.6	50.10 ± 7.6	10.5 ± 3.2	7.43 ± 6.1	12.6 ± 3.5	8.32 ± 2.9	23.1 ± 8.6	26.1 ± 11.7	1.6 ± 0.46	1.02 ± 0.13
43.27 ± 7.1	29.99 ± 8.4	14 ± 6.1	5.84 ± 4.0	14.5 ± 2.8	9.18 ± 3.6	22.65 ± 6.7	27.5 ± 5.7	1.68 ± 0.12	1.98 ± 0.51
41.57 ± 10.1	30.27 ± 10.1	14.7 ± 8.4	8.72 ± 3.6	10.3 ± 2.1	8.65 ± 2.5	21.7 ± 3.9	23.6 ± 6.8	1.87 ± 0.41	1.05 ± 0.37
43.61 ± 5.2	31.99 ± 5.9	11.6 ± 5.2	11.4 ± 4.3	12.4 ± 3.2	5.99 ± 3.1	18.69 ± 1.3	31.6 ± 4.1	1.66 ± 0.16	1.18 ± 0.20
57.96 ± 7.9	34.58 ± 4.7	15 ± 4.33	6.37 ± 1.5	16.6 ± 3.64	10.15 ± 2.0	20.64 ± 6.5	28.9 ± 3.1	1.87 ± 0.36	1.52 ± 0.13
57.43 ± 10.1	28.65 ± 9.3	12.9 ± 4.1	5.38 ± 2.7	14.5 ± 3.49	9.01 ± 0.9	26.5 ± 2.7	30.4 ± 5.6	1.6 ± 0.30	1.19 ± 0.25
60.7 ± 11.2	33.6 ± 8.5	13.4 ± 2.38	5.45 ± 1.5	13.8 ± 3.2	9.87 ± 2.3	22.73 ± 6.9	19.58 ± 6.1	1.54 ± 0.41	0.94 ± 0.39
54.9±9.02	**36.49±8.92**	**11.07±3.24**	**6.65±2.57**	**12.51±2.33**	**8.34±2.57**	**23.04±3.01**	**27.64±3.44**	**1.75±0.26**	**1.31±0.34**

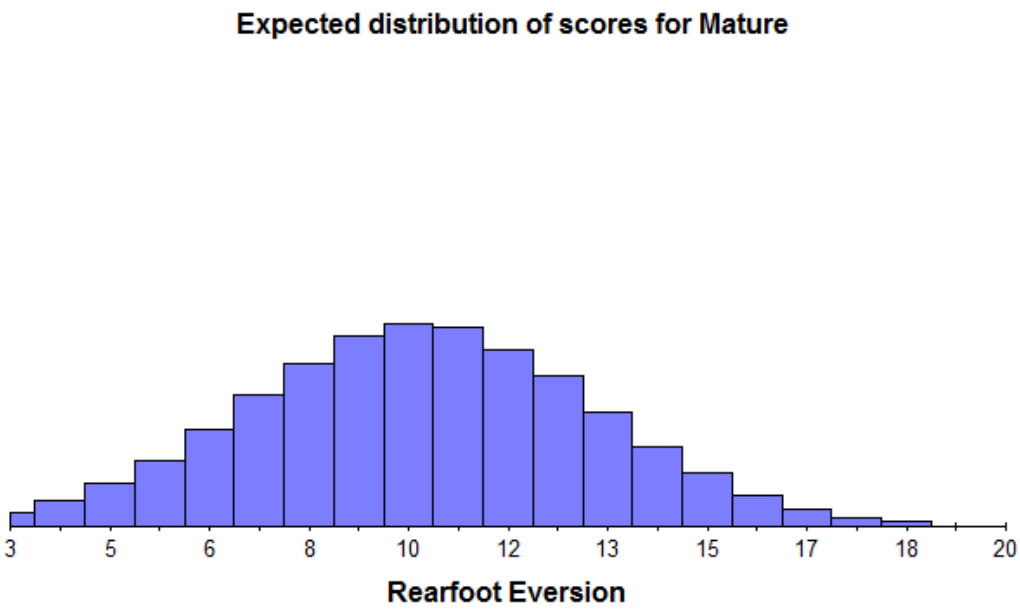
Appendix D: Sample Statistical Power Analysis report for Study 2.

Rearfoot eversion data from Study 1 was used to calculate the sample size and statistical power for study two. Calculations were performed using Power and Precision Version 4 (Borenstein, Rothstein, & Cohen, 2001).

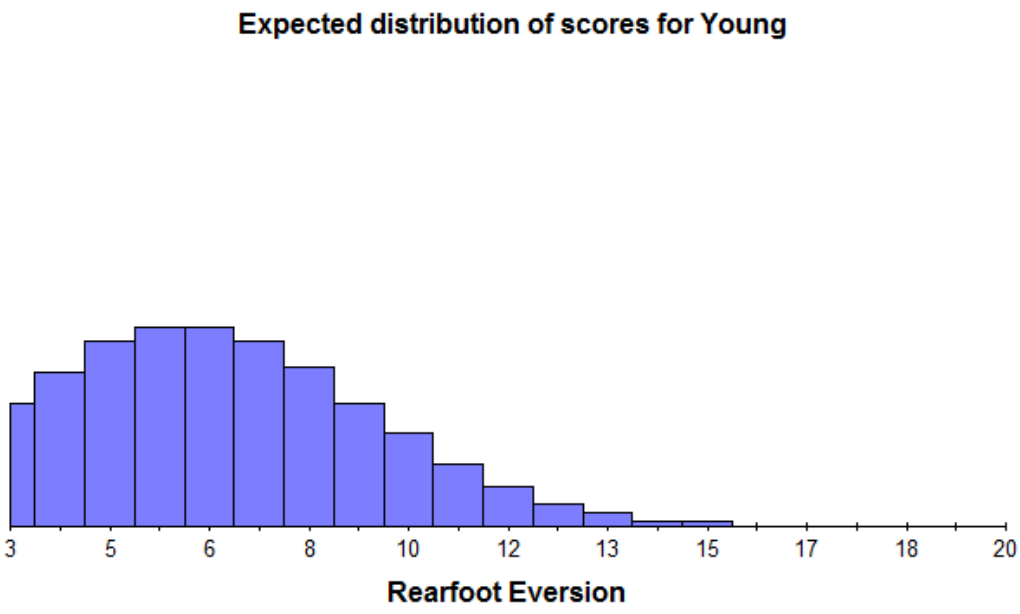
This study will compare two groups of females (mature and young) on a scale of rearfoot eversion. This variable is measured on a continuous scale with a possible range of 3 to 20 degrees. The null hypothesis is that the mean response for mature and young females is identical. The intent is therefore to disprove the null hypothesis, and show that the rearfoot eversion is significantly different for mature and young females. The computation of sample size is based on the following assumptions.

Based on the results from Study 1, it is assumed that the means produced will be 10° and 6° for the mature and young females respectively. The common within-group standard deviation was assumed to be 3°. The following graphs were calculated in the programme and illustrate the expected distribution of scores for mature (a) and young (b) females.

(a)



(b)



An expected percentage of missing data was predicted to be 10%, suggesting that for every 100 participants, 10 would provide data that could not be included in the analysis.

Results from the power analysis showed that when a difference of three degrees is predicted between the two groups (with a 10% chance of missing data), the study will have a power of 80%, and a sample size of 12 would be required to yield statistically significant results ($p<0.05$), which will allow us to conclude the groups differ.

Appendix E: Individual mean results for peak variables. Study 2.

Rearfoot Eversion.

MATURE FEMALES		YOUNG FEMALES	
Neutral	Motion Control	Neutral	Motion Control
14.1 ± 3.21	5.2 ± 1.83	17.5 ± 3.21	12 ± 2.52
21.2 ± 13.5	5.4 ± 1.78	11.4 ± 8.41	5.8 ± 3.79
22.1 ± 7.4	18.7 ± 9.94	11.8 ± 6.97	6.5 ± 1.86
26.4 ± 11.4	23.2 ± 5.62	15.3 ± 1.97	13.5 ± 5.4
5.7 ± 1.05	7 ± 2.66	11.5 ± 6.02	9.5 ± 3.49
15.9 ± 3.62	12.5 ± 2.74	12.7 ± 2.78	6.8 ± 4.23
17.1 ± 2.79	10.2 ± 5.42	17.4 ± 4.55	14.9 ± 1.84
18.1 ± 8.17	14.7 ± 6.48	14.5 ± 2.41	9.3 ± 2.9
12.3 ± 1.2	3.6 ± 1.3	7.11 ± 4.99	3.5 ± 2.01
16.4 ± 1.36	8.6 ± 3.68	4.4 ± 1.08	6.9 ± 3.61
		12.0 ± 4.07	8.9 ± 2.34
		9.12 ± 3.94	8.36 ± 2.77
17.0 ± 5.6	**10.91 ± 4.1**	**12.06 ± 3.9**	**8.8 ± 3.3**

Ankle Dorsiflexion.

MATURE FEMALES		YOUNG FEMALES	
Neutral	Motion Control	Neutral	Motion Control
29.1 ± 5.67	27.7 ± 6.93	52 ± 17.4	42 ± 17.1
23.9 ± 4.89	27.7 ± 7.7	30.5 ± 8.03	38 ± 15.6
31.1 ± 12.4	22.8 ± 5.18	25.4 ± 2.03	26.4 ± 1.32
34.5 ± 7.67	30.2 ± 8.7	20.1 ± 10.8	18.1 ± 10.6
22.3 ± 2.9	26.6 ± 7.5	55.2 ± 18	31.4 ± 8.1
25 ± 4.26	21.4 ± 5.57	30.8 ± 5.77	31.4 ± 7.47
10.9 ± 1.68	26.2 ± 10.23	18.6 ± 4.4	18 ± 4.2
27.2 ± 8.9	20.6 ± 9.56	33.9 ± 5.6	34.3 ± 2.67
29.8 ± 5.69	31.7 ± 7.35	30.3 ± 1.11	45.5 ± 4.22
28.1 ± 7.51	27.3 ± 9.91	42.9 ± 11.5	64.4 ± 13.5
		25 ± 1.83	29.8 ± 2.29
		41.6 ± 2.08	31.9 ± 2.9
26.2 ± 6.5	**26.2 ± 3.6**	**33.9 ± 12.6**	**34.3 ± 12.6**

Knee Flexion.

MATURE FEMALES		YOUNG FEMALES	
Neutral	Motion Control	Neutral	Motion Control
33.9 ± 3.7	31.4 ± 2.6	47.6 ± 6.8	47.9 ± 3.7
13.2 ± 2.6	19 ± 1.9	17.4 ± 6.0	24 ± 8.14
31.2 ± 5.9	31.8 ± 7.2	16.6 ± 2.53	14.6 ± 3.0
43.3 ± 12.9	35.7 ± 7.9	26.9 ± 8.0	28.1 ± 2.7
20 ± 1.45	35.2 ± 2.04	22.2 ± 9.0	33.3 ± 6.9
42.1 ± 12.0	29.1 ± 7.2	34.4 ± 5.7	31.6 ± 3.1
36.6 ± 14.36	17.4 ± 4.9	29.8 ± 2.0	27.6 ± 3.0
44.8 ± 11.0	44.0 ± 1.2	32.5 ± 1.8	34.3 ± 4.9
18.2 3.2	46.1 ± 7.6	6.2 ± 8.0	4.1 ± 4.0
39.6 ±2.0	32.7 ± 6.0	28.0 ± 2.7	30.5 ± 2.66
		41.4 ± 4.0	44 ± 6.0
		43.7 ± 5.1	46.2 ± 5.0
32.3 ± 11.4	**32.2 ± 9.2**	**28.9 ± 12.12**	**30.5 ± 12.65**

Knee Abduction.

MATURE FEMALES		YOUNG FEMALES	
Neutral	Motion Control	Neutral	Motion Control
24.7 ± 3.65	28.6 ± 7.69	20.6 ± 5.01	28.6 ± 3.2
27.9 ± 3.92	29.7 ± 1.924	14.7 ± 1.59	29.7 ± 6.75
24 ± 10.3	24.1 ± 4.05	11.1 ± 5.49	11.1 ± 6.88
21.1 ± 2.85	23.3 ± 1.96	23.3 ± 3.65	23.3 ± 3.0
20.4 ± 3.26	20.0 ± 5.91	10.0 ± 5.89	10.0 ± 3.87
32.35 ± 5.11	34.6 ± 8.43	25.1 ± 2.71	24.6 ± 3.17
27.3 ± 9.63	28.9 ± 10.54	23.9 ± 7.2	26.6 ± 6.3
20.0 ± 5.8	21.0 ± 1.89	21.0 ± 3.9	21.0 ± 7.9
17.2 ± 4.54	13.5 ± 7.54	13.5± 8.43	13.5 ± 8.09
16.1 ± 2.0	16.7 ± 1.13	6.7 ± 3.23	6.7 ± 7.6
		16.0 ± 6.49	15.9 ± 6.22
		18.0 ± 3.9	18.1 ± 3.58
23.1 ± 5.1	**24.0 ± 6.5**	**17.0 ± 5.96**	**17.1 ± 6.38**

Knee Internal Rotation.

MATURE FEMALES		YOUNG FEMALES	
Neutral	Motion Control	Neutral	Motion Control
16.0 ± 3.2	16 ± 2.53	13.8 ± 2.21	10.3 ± 2.54
11.6 ± 3.90	11.6 ± 2.67	15.5 ± 3.53	12.5 ± 3.07
18.3 ± 5.9	18.3 ± 7.2	13.5 ± 2.75	10 ± 2.61
19.3 ± 3.15	19.3 ± 7.3	12.7 ± 3.3	12.8 ± 6.9
19.4 ± 2.81	19.4 ± 4.8	16 ± 2.58	11.91 ± 3.641
10.8 ± 3.28	10.8 ± 3.76	18.3 ± 2.3	18.2 ± 1.94
14.0 ± 3.42	14 ± 2.89	10.2 ± 2.36	13.8 ± 2.69
14.8 ± 1.53	14.8 ± 7.5	13.9 ± 2.6	11.65 ± 3.3
12.6 ± 2.75	12.6 ± 2.61	15.4 ± 6.5	13.4 ± 7.1
14.5 ± 3.15	14.5 ± 3.18	16 ± 3.2	16 ± 3.96
		12 ± 1.8	9 ± 2.4
		14.0 ± 2.09	11.9 ± 2.76
18.4 ± 3.7	**15.1 ± 3.1**	**14.3 ± 2.14**	**12.6 ± 2.56**

Ground Reaction Force Results.

Mature Females.

MATURE FEMALES Mean (SD)		VARIABLE			
		Peak Impact Force (BW)	Occurrence Time (PkFz)	Loading Rate (BW)	Occurrence Time (LR)
1	N	1.82 (0.5)	0.024 (0.007)	90.964 (21.934)	0.021 (0.004)
	MC	1.84 (0.34)	0.025 (0.004)	79.054 (15.564)	0.020 (0.006)
2	N	2.45 (0.79)	0.032 (0.005)	94.685 (31.813)	0.025 (0.006)
	MC	2.34 (0.77)	0.031 (0.004)	103.53 (33.77)	0.028 (0.005)
3	N	2.11 (0.52)	0.025 (0.006)	90.253 (7.6459)	0.018 (0.009)
	MC	1.95 (0.56)	0.026 (0.004)	101.05 (25.1)	0.020 (0.003)
4	N	1.78 (0.95)	0.017 (0.005)	122.52 (14.497)	0.024 (0.002)
	MC	1.79 (0.64)	0.019 (0.004)	82.372 (6.35)	0.023 (0.002)
5	N	1.98 (0.45)	0.024 (0.006)	173.25 (22.09)	0.017 (0.005)
	MC	1.89 (0.5)	0.028 (0.004)	186 (22.89)	0.019 (0.006)
6	N	2.22 (0.09)	0.030 (0.005)	118 (13.2)	0.016 (0.006)
	MC	2.23 (0.10)	0.029 (0.004)	125.36 (20.3)	0.016 (0.005)
7	N	1.99 (0.29)	0.034 (0.002)	77.59 (7.9)	0.019 (0.003)
	MC	2.08 (0.30)	0.028 (0.001)	76.77 (6.23)	0.021 (0.006)
8	N	2.97 (0.55)	0.018 (0.006)	100.36 (5.4)	0.025 (0.007)
	MC	2.94 (0.67)	0.021 (0.006)	102.31 (5.55)	0.026 (0.002)
9	N	1.70 (0.89)	0.017 (0.003)	112.35 (19.5)	0.028 (0.005)
	MC	2.16 (0.59)	0.019 (0.002)	98.23 (21.2)	0.029 (0.003)
10	N	1.23 (0.3)	0.026 (0.004)	90.65 (13.97)	0.015 (0.006)
	MC	1.51 (0.28)	0.027 (0.005)	125.36 (13.25)	0.017 (0.004)
Mean (SD)	**N**	**2.07 (0.39)**	**0.025 (0.006)**	**107.06 (21.17)**	**0.021 (0.004)**
	MC	**2.03 (0.47)**	**0.025 (0.004)**	**108.06 (32.24)**	**0.022 (0.004)**

Young Females.

YOUNG FEMALES Mean (SD)		VARIABLE			
		Peak Impact Force (BW)	Occurrence Time (PkFz)	Loading Rate (BW)	Occurrence Time (LR)
1	N	1.93 (0.69)	0.019 (0.004)	156.54 (22.09)	0.021 (0.007)
	MC	2.21 (0.6)	0.023 (0.004)	186.62 (22.89)	0.024 (0.003)
2	N	2.45 (0.7)	0.021 (0.003)	204.9 (32.687)	0.024 (0.002)
	MC	2.21 (0.8)	0.024 (0.007)	266.3 (59.18)	0.023 (0.002)
3	N	2.26 (0.55)	0.027 (0.005)	148.3 (66.453)	0.015 (0.006)
	MC	1.66 (0.94)	0.022 (0.004)	163.8 (32.644)	0.019 (0.007)
4	N	2.06 (0.089)	0.028 (0.007)	86.5 (17.52709)	0.016 (0.004)
	MC	2.02 (0.10)	0.027 (0.006)	79.3 (7.6308)	0.019 (0.006)
5	N	1.59 (0.1)	0.029 (0.004)	100.61 (10.298)	0.018 (0.006)
	MC	2.29 (0.24)	0.022 (0.003)	117.4 (25.91)	0.020 (0.003)
6	N	2.37 (0.36)	0.027 (0.005)	66.51 (10.781)	0.019 (0.001)
	MC	2.17 (0.49)	0.026 (0.005)	74.3 (6.8011)	0.018 (0.002)
7	N	2.34 (0.6)	0.023 (0.007)	168.21 (20.681)	0.026 (0.003)
	MC	2.39 (0.57)	0.024 (0.004)	187.1 (26.261)	0.026 (0.004)
8	N	2.76 (0.80)	0.020 (0.006)	149.4 (29.595)	0.019 (0.004)
	MC	2.68 (0.98)	0.019 (0.001)	138.9 (60.94)	0.019 (0.003)
9	N	2.13 (0.80)	0.018 (0.006)	125.52 (47.189)	0.023 (0.005)
	MC	2.31 (0.80)	0.021 (0.003)	117.3 (24.75)	0.026 (0.004)
10	N	2.47 (0.65)	0.029 (0.005)	61.453 (21.218)	0.025 (0.005)
	MC	2.32 (0.45)	0.018 (0.009)	75.745 (15.128)	0.028 (0.005)
11	N	2.53 (0.83)	0.026 (0.002)	73.439 (7.3917)	0.025 (0.004)
	MC	2.54 (0.75)	0.025 (0.004)	82.122 (9.3442)	0.026 (0.003)
12	N	1.9 (0.22)	0.024 (0.001)	84.572 (5.84)	0.028 (0.004)
	MC	2.21 (0.35)	0.027 (0.005)	85.444 (6.9035)	0.026 (0.002)
Mean (SD)	**N**	**2.36 (0.23)**	**0.024 (0.004)**	**126.79 (47.01)**	**0.021 (0.004)**
	MC	**2.25 (0.25)**	**0.023 (0.003)**	**140.68 (61.52)**	**0.022 (0.004)**

Appendix F. Individual mean results from Study 3 for peak values in each footwear condition.

Rearfoot Eversion.

PARTICIPANT	REARFOOT EVERSION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	9.8 ± 1.15	4.8 ± 1.2	6.2 ± 0.85	10.32 ± 1.33
2	14.4 ± 4.38	8.4 ± 2.61	12.9 ± 2.67	11.25 ± 2.37
3	14.5 ± 3.9	9.75 ± 1.63	15.3 ± 4.34	17.2 ± 5.61
4	19.9 ± 2.67	12.33 ± 2.1	16.5 ±1.7	16.5 ± 6.3
5	13.1 ± 6.42	7.8 ± 4.46	13.2 ± 5.72	15.3 ± 4.03
6	18.5 ± 5.7	13.2 ± 3.6	17.5 ± 5.26	18 ± 3.79
7	19.6 ± 3.22	10.2 ± 2.5	19.8 ± 6.59	20.9 ± 5.89
8	14.2 ± 2.29	7.43 ± 1.78	13.7 ± 2.861	14.6 ± 0.41
9	12.6 ± 1.2	4.89 ± 5.4	8.7 ± 2.6	14.9 ± 1.84
10	7.17 ± 3.44	5.16 ± 1.72	6.56 ± 5.5	6.69 ± 1.57
11	13.8 ± 4.05	8.4 ± 2.77	11.5 ± 4.51	13.9 ± 3.6
12	9.8 ± 3.25	5.6 ± 6.9	7.8 ± 3.23	11.4 ± 3.27
13	17.2 ± 2.82	10.2 ± 1.53	19.6 ± 2.84	20.4 ± 2.9
14	10.3 ± 1.57	3.8 ± 2.6	10.6 ± 5.58	13.2 ± 3.84
15	10.56 ± 1.94	13.56 ± 1.15	20.8 ± 2.82	24.6 ± 1.08
16	16.4 ± 8.5	8.33 ± 2.74	20.6 ± 3.22	22.5 ± 3.22
17	14.7 ± 1.107	9.23 ± 3.23	13.7 ± 2.7	12.9 ± 3.23
18	14.05 ± 4.4	8.95 ± 2.6	10.3 ± 2.84	15.002 ± 1.87
19	14.64 ± 3.13	9.05 ± 4.63	13.5 ± 1.8	17.5 ± 2.5
20	15.36 ± 1.28	8.4 ± 1.3	14.3 ± 1.6	16.8 ± 1.64
Mean (SD)	**14.03 ± 3.7**	**8.47 ± 4.2**	**13.65 ± 4.9**	**15.69 ± 4.68**

Ankle Dorsiflexion.

PARTICIPANT	ANKLE DORSIFLEXION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	20.5 ± 3.5	19.2 ± 1.14	24.6 ± 4.6	19.9 ± 2.33
2	20 ± 3.11	25.6 ± 2.83	19.5 ±5.35	19.3 ±3.3
3	25 ± 5.2	20.4 ± 5.69	20 ± 4.13	20.9 ± 5.078
4	20.3 ± 5.31	18.9 ± 5.1	18.2 ± 5.58	19.1 ± 5.14
5	16.4 ± 6.3	13.5 ± 4.4	13.4 ± 2.24	13.8 ± 2.03
6	23.3 ± 1.99	17.2 ± 2.4	16.2 ± 2.22	16.4 ± 1.27
7	11.3 ± 3.6	11.5 ± 4.16	12.5 ± 5.6	11.4 ± 5.33
8	25.7 ± 7.36	20.5 ± 6.83	19.4 ± 1.35	18.4 ± 1.82
9	18.5 ± 2.81	20.5 ± 1.22	18.61 ± 3.397	21 ± 5.112
10	21 ± 1.3	20.3 ± 1.18	19.23 ± 1.85	21.7 ± 1.57
11	16.5 ± 2.69	13.2 ± 1.08	13.4 ± 1.56	10.2 ± 1.3
12	16.5 ± 1.995	18.3 ± 2.357	20.3 ± 7.13	20.4 ± 5.9
13	24.5 ± 10.07	21.3 ± 9.09	24.4 ± 5.73	23.5 ± 9.56
14	22 ± 1.13	17.8 ± 2.15	20.7 ± 2.38	20.6 ± 1.98
15	24.2 ± 1.91	21.1 ± 1.09	20.1 ± 1.79	26.4 ± 1.11
16	17 ± 1.7	14.2 ± 1.1	15.1 ± 1.62	16.5 ± 1.9
17	19.9 ± 4.49	16.2 ± 4.75	21.4 ± 6.97	16.5 ± 5.18
18	16 ± 1.35	12.5 ± 2.2	13.5 ± 3.2	18.9 ± 3.93
19	18.9 ± 4.9	13.9 ± 3.5	15.2 ± 2.6	17.4 ± 1.5
20	19.5 ± 1.4	17.8 ± 2.4	17 ± 6.45	19.2 ± 3.6
Mean (SD)	**19.9 ± 3.64**	**17.7 ± 1.83**	**18.2 ± 3.54**	**18.8 ± 3.88**

Forefoot Abduction.

PARTICIPANT	FOREFOOT ABDUCTION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	19.2 ± 1.65	20.5 ± 0.74	24.6 ± 2.03	19.9 ± 2.04
2	10.92 ± 3.5	25.81 ± 2.99	6.19 ± 6.8	20.58 ± 3.9
3	9.6 ± 5.221	13.8 ± 4.09	10.5 ± 6.23	9.54 ± 5.32
4	20.1 ± 3.17	17.16 ± 3.7	36 ± 2.65	29.5 ± 3.67
5	21.2 ± 1.42	28.1 ± 1.878	26.6 ± 2.936	26.1± 3.82
6	18.4 ± 3.950	11.3 ± 3.875	15.96 ± 4.137	15.39 ± 4.13
7	14.2 ± 3.25	11.8 ± 3.002	12.6 ± 1.05	13.2 ± 0.96
8	12.1 ± 2.34	5.6 ± 1.51	11.9 ± 3.48	11.8 ± 1.86
9	8.5 ± 1.745	7.5 ± 3.26	10.5 ± 2.81	9.7 ± 2.64
10	7.64 ± 4.69	6.66 ± 4.32	8.26 ± 5.67	8.06 ± 4.17
11	10.9 ± 6.9	4.12 ± 2.67	5.79 ± 3.63	3.54 ± 2.757
12	5.42 ± 3.07	3.55 ± 5.08	7.41 ± 3.08	6.33 ± 4.69
13	4.91 ± 2.73	3.47 ± 1.92	5.68 ± 4.73	3.95 ± 4.20
14	7.26 ± 3.8	3.07 ± 5.6	5.08 ± 1.374	4.69 ± 2.08
15	8.02 ± 6.19	7.99 ± 2.17	5.6 ± 9.05	6.87 ± 7.01
16	11.7 ± 3.4	11.2 ± 2.19	7.49 ± 1.5	8.94 ± 5.87
17	8.6 ± 6.35	8.14 ± 6.0	3.08 ± 2.8	3.07 ±6.6
18	11.8 ± 6.9	11.1 ± 4.2	13.2 ± 2.72	13.3 ± 3.32
19	11 ± 3.5	10.9 ± 3.07	10.5 ± 2.4	10.2 ± 2.8
20	13.2 ± 3.45	12 ± 4.24	11.95 ± 3.08	11.9 ± 3.93
Mean (SD)	**11.7 ± 5.18**	**11.2 ± 7.69**	**11.95 ± 9.02**	**11.83 ± 7.97**

Knee Flexion.

PARTICIPANT	KNEE FLEXION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	27 ± 1.59	20 ± 1.55	25 ± 2.07	24 ± 1.66
2	31.6 ± 5.7	48.7 ± 5.3	30.4 ± 2.9	31.2 ± 6.2
3	24 ± 4.52	37 ± 5.876	23.7 ± 4.6	20.1 ± 3.54
4	35 ± 4.1	32 ± 4.5	32 ± 4.05	33 ± 3.71
5	44.5 ± 5.4	40.2 ± 3.3	38.4 ± 5.0	38.8 ± 3.2
6	46.2 ± 5.8	37.5 ± 5.9	38.7 ± 4.9	29.7 ± 3.4
7	38 ± 3.41	34 ± 2.6971	34 ± 3.01	33 ± 5.99
8	35 ± 3.2	23 ± 4.1	33 ± 7.5	31 ± 4.6
9	39 ± 3.02	43 ± 2.23	45 ± 3.14	43 ± 2.406
10	33 ± 5.5	31 ± 3.2	31 ± 6.5	31 ± 4.9
11	36 ± 2.6	30 ± 3.3	33 ± 1.86	44 ± 1.45
12	28 ± 4.09	29 ± 2.726	28 ± 1.2	29 ± 5.3
13	40 ± 1.57	34 ± 3.9	34 ± 3.9	33 ± 7.0
14	36 ± 2.8	35 ± 4.56	34 ± 3.9	33 ± 5.3
15	41 ± 4.8	45 ± 3.1	42 ± 2.2	40 ± 2.1
16	36 ± 3.7	35 ± 3.0	37 ± 4.7	31 ± 4.0
17	29 ± 7.8	30 ± 4.7	31 ± 4.3	29 ± 1.6
18	36.2 ± 5.4	35.8 ± 3.4	33 ± 5.9	32 ± 5.4
19	35 ± 3.96	32.4 ± 2.5	31.6 ± 1.79	33.6 ± 1.7
20	34.5 ± 1.1	34.9 ± 3.7	35.9 ± 1.7	32.2 ± 4.3
Mean (SD)	**35.25 ± 5.58**	**34.38 ± 6.77**	**33.54 ± 5.14**	**32.58 ± 5.66**

Knee Abduction.

PARTICIPANT	KNEE ABDUCTION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	10.2 ± 4.11	6.69 ± 2.31	13.1 ± 0.94	6.72 ± 1.07
2	16.9 ± 1.381	10.1 ± 1.15	15.08 ± 4.3	7.7 ± 1.3
3	25.1 ± 1.73	23.6 ± 2.7	25.2 ± 2.14	25.5 ± 1999
4	5.6 ± 4.15	5.4 ± 3.06	8.9 ± 3.63	4.1 ± 3.64
5	36.2 ± 1.23	20 ± 1.1589	20.8 ± 3.83	20.1 ± 5.54
6	8.1 ± 6.48	7.9 ± 4.4	10.2 ± 1.79	8.1 ± 1.59
7	12.5 ± 2.65	7.9 ± 1.39	13 ± 4.06	9.8 ± 1.33
8	13.65 ± 5.5	2.5 ± 1.13	11.2 ± 1.9	10 ± 1.81
9	11.1 ± 5.3	8.8 ± 3.316	11.3 ± 3.4	9.1 ± 3.3
10	12.79 ± 6.1	10.87 ± 4.1	12.4 ± 3.93	12.4 ± 4.72
11	14.52 ± 3.94	4.09 ± 2.59	7 ± 1.82	5.77 ± 3.9
12	18.1 ± 2.4	16.6 ± 3.5	16.9 ± 6.3	16.8 ± 2.49
13	8.85 ± 1.12	6.33 ± 3.63	10.81 ± 2.62	5.69 ± 2.667
14	3.4 ± 1.31	3.5 ± 2.6	5.99 ± 1.32	4.089 ± 2.89
15	3.9365 ± 1.59	2.65 ± 1.93	3 ± 0.76	1.1 0.71
16	12.3 ± 0.7	9.5 ± 3.9	10.9 ± 5.24	7.7 ± 6.5
17	9.9 ± 1.71	16.9 ± 1.71	13.9 ± 1.38	10.86 ± 0.96
18	10.5 ± 2.4	9.2 ± 3.0	9.8 ± 2.5	9 ± 2.9
19	12.8 ± 5.39	11.2 ± 4.13	12.9 ± 3.7	10.2 ± 3.4
20	16.2 ± 1.99	11.2 ± 1.7	13.9 ± 3.3	9.2 ± 3.7
Mean (SD)	**13.13 ± 7.37**	**9.74 ± 5.72**	**12.31 ± 4.91**	**9.69 ± 5.65**

Knee Internal Rotation.

PARTICIPANT	KNEE INTERNAL ROTATION			
	NEUTRAL	ORTHOTICS	MEDIAL WEDGE	LATERAL WEDGE
1	11.4 ± 0.98	10.6 ± 1.46	11.6 ± 2.04	13.1 ± 0.99
2	14.6 ± 4.38	4.5 ± 2.61	7.6 ± 1.14	16.3 ± 2.13
3	17.4 ± 1.23	10.2 ± 1.89	11.3 ± 1.18	20.3 ± 2.03
4	14.3 ± 2.3	9.1 ± 7.9	14.1 ± 1.23	14.0 ± 1.3
5	12.5 ± 1.91	9.2 ± 1.41	12.0 ± 2.03	12.3 ± 1.1
6	9.5 ± 1.1	6.2 ± 1.8	8.8 ± 1.78	8.5 ± 1.48
7	12 ± 3.99	7.9 ± 3.59	14.4 ± 4.29	13.8 ± 3.65
8	14.2 ± 1.92	11.3 ± 3.7	13.6 ±1.24	15.2 ± 2.29
9	14.7 ± 1.56	13.8 ± 1.0	14.2 ± 1.82	12.7 ± 1.68
10	15 ± 2.94	12.6 ± 3.3	14.6 ± 3.05	15.1 ± 2.42
11	14 ± 2.59	11.3 ± 3.51	16 ± 4.77	12.8 ± 4.71
12	20.5 ± 3.0	16.8 ± 1.84	18.5 ± 2.8	20.4 ± 3.3
13	14 ± 2.4	9.2 ± 2.12	18.9 ± 2.4	17.2 ± 2.04
14	13.9 ± 2.72	7.2 ± 4.09	8.9 ± 3.29	10.1 ± 3.10
15	11.2 ± 1.01	8.6 ± 2.7	12.2 ± 2.9	10.5 ± 1.9
16	15 ± 5.3	9 ± 2.79	12.2 ± 2.8	13.8 ± 3.0
17	14.6 ± 1.76	11 ± 1.74	11.2 ± 1.8	15.5 ± 3.1
18	16.2 ± 2.0	8.22 ± 2.6	12 ± 2.4	13 ± 1.4
19	16.3 ± 3.99	9.75 ± 3.05	14.22 ± 3.5	14.2 ± 3.8
20	17.1 ± 4.2	9.89 ± 1.7	13.6 ± 2.3	13.9 ± 1.9
Mean (SD)	**14.4 ± 2.47**	**9.81 ± 2.68**	**12.99 ± 2.84**	**14.13 ± 3.06**

Knee External Adductor Moment and Moment Arm.

PARTICIPANT	FOOTWEAR CONDITION Nm – Nm/kg			
	NEUTRAL	***ORTHOTIC***	***MEDIAL***	***LATERAL***
1	74.7495 ± 6.25 **1.23** ± 0.10	30.12627 ± 3.2 **0.55** ± 0.053	38.30268 ± 4.5 **0.70** ± 0.075	33.7389 ± 3.55 **0.61** ± 0.059
2	46.3969 ± 4.11 **0.89** ± 0.079	36.24614 ± 5.66 **0.70** ± 0.11	45.32104 ± 4.95 **0.87** ± 0.095	45.07996 ± 4.32 **0.87** ± 0.083
3	49.247 ± 5.06 **0.93** ± 0.095	27.07696 ± 4.06 **0.51** ± 0.077	51.44997 ± 5.55 **0.97** ± 0.10	58.17689 ± 6.54 **1.10** ± 0.12
4	72.54 ± 9.55 **1.27** ± 0.17	50.04 ± 9.16 **0.88** ± 0.16	52.25479 ± 8.5 **0.92** ± 0.15	51.7093 ± 9.43 **0.91** ± 0.17
5	65.6718 ± 5.33 **1.22** ± 0.099	38.43344 ± 6.03 **0.71** ± 0.11	57.41337 ± 7.41 **1.06** ± 0.14	47.017 ± 5.02 **0.87** ± 0.093
6	68.4206 ± 5.56 **1.37** ± 0.11	52.12537 ± 4.89 **1.04** ± 0.098	58.23274 ± 5.12 **1.16** ± 0.10	58.57561 ± 6.96 **1.17** ± 0.14
7	52.587 ± 4.52 **1.03** ± 0.089	39.2218 ± 6.2 **0.77** ± 0.12	57.3875 ± 7.46 **1.13** ± 0.15	47.5041 ± 5.12 **0.93** ± 0.10
8	59.2742 ± 6.33 **1.16** ± 0.12	46.63385 ± 6.0 **0.91** ± 0.12	31.4967 ± 5.73 **0.62** ± 0.11	49.9547 ± 5.16 **0.98** ± 0.10
9	58.265 ± 5.98 **1.17** ± 0.12	49.5254 ± 5.33 **0.99** ± 0.11	59.625 ± 5.0 **1.19** ± 0.10	59.0125 ± 5.91 **1.18** ± 0.12
10	64.254 ± 7.15 **1.11** ± 0.12	64.256 ± 5.41 **1.11** ± 0.093	66.662 ± 8.55 **1.15** ± 0.15	65.258 ± 9.41 **1.13** ± 0.16
11	66.958 ± 6.25 **1.24** ± 0.12	60.235 ± 6.68 **1.12** ± 0.12	68.254 ± 7.15 **1.26** ± 0.13	64.0333 ± 6.55 **1.19** ± 0.12
12	70.152 ± 9.51 **1.19** ± 0.16	67.198 ± 8.07 **1.14** ± 0.14	74.558 ± 9.13 **1.26** ± 0.15	72.0214 ± 9.55 **1.22** ± 0.16
13	59.215 ± 4.87 **1.04** ± 0.085	54.32 ± 5.96 **0.95** ± 0.10	55.625 ± 5.81 **0.98** ± 0.10	55.0541 ± 3.22 **0.97** ± 0.056
14	71.223 ± 5.91 **1.19** ± 0.099	72.354 ± 8.01 **1.2** ± 0.12	69.568 ± 9.66 **1.16** ± 0.16	70.113 ± 9.84 **1.17** ± 0.16
15	69.35 ± 9.22 **1.24** ± 0.16	58.889 ± 9.84 **1.05** ± 0.18	66.691 ± 10.82 **1.19** ± 0.19	68.554 ± 10.03 **1.22** ± 0.18
16	60.325 ± 4.03 **1.10** ± 0.073	59.39636 ± 3.23 **1.08** ± 0.059	58.85 ± 2.16 **1.07** ± 0.039	56.365 ± 3.08 **1.02** ± 0.056
17	60.254 ± 9.66 **1.04** ± 0.17	49.14 ± 8.84 **0.85** ± 0.15	60.39 ± 7.16 **1.04** ± 0.12	55.254 ± 9.0 **0.95** ± 0.16
18	55.214 ± 4.69 **1.13** ± 0.096	50.23146 ± 5.67 **1.03** ± 0.12	53.265 ± 6.04 **1.09** ± 0.12	52.325 ± 5.01 **1.07** ± 0.10
19	60.32 ± 9.13 **1.14** ± 0.17	47.78 ± 9.84 **0.90** ± 0.19	60.98 ± 9.64 **1.15** ± 0.18	51.236 ± 7.89 **0.97** ± 0.15
20	60.258 ± 8.46 **1.12** ± 0.16	46.3589 ± 8.61 **0.86** ± 0.16	57.856 ± 7.11 **1.07** ± 0.13	49.826 ± 8.64 **0.92** ± 0.16
MEAN	**62.23** **1.15**	**49.80*** **0.92***	**57.21** **1.05**	**55.54*** **1.02***
STANDARD DEVIATION	**7.76** **0.12**	**11.89** **0.19**	**10.32** **0.17**	**9.40** **0.15**

*Differences were statistically significant compared to neutral condition at $p<0.05$ level.

PARTICIPANTS	NEUTRAL		ORTHOTIC		MEDIAL		LATERAL	
	LATERAL		*LATERAL*		*LATERAL*		*LATERAL*	
	Mean	**Occurrence Time**	**Mean**	**Occurrence Time**	**Mean**	**Occurrence Time**	**Mean**	**Occurrence Time**
1	-0.021 ± 0.011	0.22 ± 0.009	-0.019 ± 0.004	0.18 ± 0.013	-0.018 ± 0.003	0.20 ± 0.014	-0.016 ± 0.005	0.17 ± 0.015
2	-0.019 ± 0.008	0.21 ± 0.005	-0.017 ± 0.01	0.19 ± 0.009	-0.018 ± 0.007	0.21 ±0.016	-0.013 ± 0.006	0.19 ± 0.016
3	-0.019 ± 0.009	0.18 ± 0.02	-0.016 ± 0.003	0.19 ± 0.014	-0.020 ± 0.013	0.19 ± 0.01	-0.02 ± 0.009	0.20 ± 0.01
4	-0.020 ± 0.009	0.19 ± 0.01	-0.018 ± 0.009	0.21 ± 0.007	-0.021 ± 0.0091	0.20 ± 0.013	-0.015 ± 0.009	0.20 ± 0.009
5	-0.020 ± 0.012	0.18 ± 0.01	-0.017 ± 0.007	0.19 ± 0.012	-0.022 ± 0.005	0.20 ± 0.01	-0.019 ± 0.005	0.21 ± 0.016
6	-0.023 ± 0.013	0.21 ± 0.015	-0.020 ± 0.006	0.18 ± 0.009	-0.022 ± 0.01	0.21 ± 0.015	-0.021 ± 0.013	0.21 ± 0.013
7	-0.018 ± 0.11	0.22 ± 0.016	-0.019 ± 0.009	0.19 ± 0.01	-0.002 ± 0.005	0.24 ± 0.007	-0.017 ± 0.011	0.20 ± 0.01
8	-0.020 ± 0.008	0.18 ± 0.017	-0.020 ± 0.005	0.19 ± 0.011	-0.014 ± 0.008	0.18 ± 0.009	-0.016 ± 0.009	0.21 ± 0.01
9	-0.020 ± 0.010	0.22 ± 0.009	-0.020 ± 0.001	0.21 ± 0.01	-0.023 ± 0.002	0.19 ± 0.01	-0.021 ± 0.007	0.18 ± 0.015
10	-0.019 ± 0.012	0.19 ± 0.009	-0.019 ± 0.001	0.22 ± 0.009	-0.013 ± 0.004	0.20 ± 0.009	-0.023 ± 0.01	0.16 ± 0.013
11	-0.020 ± 0.009	0.21 ± 0.2	-0.018 ± 0.001	0.19 ± 0.008	-0.021 ± 0.006	0.19 ± 0.009	-0.02 ± 0.006	0.18 ± 0.016
12	-0.020 ± 0.013	0.20 ± 0.008	-0.020 ± 0.0012	0.19 ± 0.01	-0.019 ± 0.011	0.19 ± 0.01	-0.021 ± 0.003	0.17 ± 0.016
13	-0.019 ± 0.021	0.18 ± 0.010	-0.019 ± 0.009	0.18 ± 0.01	-0.018 ± 0.009	0.19 ± 0.011	-0.02 ± 0.013	0.16 ± 0.02
14	-0.018 ± 0.003	0.18 ± 0.010	-0.02 ± 0.005	0.22 ± 0.014	-0.021 ± 0.009	0.21 ± 0.011	-0.02 ± 0.006	0.17 ± 0.016
15	-0.020 ± 0.011	0.19 ± 0.007	-0.017 ± 0.003	0.19 ± 0.01	-0.022 ± 0.007	0.20 ± 0.009	-0.017 ± 0.009	0.20 ± 0.011
16	-0.19 ± 0.002	0.19 ± 0.013	-0.018 ± 0.009	0.19 ± 0.009	-0.023 ± 0.0012	0.20 ± 0.01	-0.017 ± 0.004	0.20 ± 0.009
17	-0.018 ± 0.011	0.20 ± 0.010	-0.016 ± 0.005	0.20 ± 0.008	-0.022 ± 0.003	0.20 ± 0.008	-0.018 ± 0.003	0.21 ± 0.009
18	-0.021 ± 0.0013	0.21 ± 0.011	-0.020 ± 0.001	0.21 ± 0.012	-0.017 ± 0.005	0.21 ± 0.013	-0.02 ± 0.007	0.19 ± 0.01
19	-0.021 ± 0.007	0.22 ± 0.012	-0.017 ± 0.008	0.21 ± 0.012	-0.018 ± 0.004	0.17 ± 0.019	-0.020 ± 0.007	0.19 ± 0.008
20	-0.020 ± 0.016	0.23 ± 0.008	-0.020 ± 0.0012	0.18 ± 0.01	-0.018 ± 0.12	0.17 ± 0.016	-0.015 ± 0.008	0.18 ± 0.017
Mean (SD)	**-0.020** ± 0.001	**0.20** ± 0.016	**-0.019** ± 0.001	**0.20** ± 0.013	**0.020** ± 0.003	**0.20** ± 0.016	**0.018** ± 0.003	**0.19** ± 0.017

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