THE EFFECTS OF ORTHOTICS ON LOWER EXTREMITY VARIABILITY DURING RUNNING.

By

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Submitted to the graduate degree program in Health, Sport, and Exercise Science and the Graduate Faculty of the University of Kansas in partial fulfillment of the requirements for the degree of Master of Science.

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Date Defended: 12/07/2011

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Date approved: 12/07/2011

Abstract

Introduction: Abnormal foot mechanics may affect kinematics of the lower extremity, predisposing individuals to injury. Foot orthotics are often used to alter lower extremity mechanics. Little research is available examining effects of orthotics on lower extremity kinematic variability during running. **Methods**: 30 recreational runners (15 males, 15 females) identified as excessive pronators participated in this study. Subjects were tested with and without orthotics while running on a treadmill at 3.35 m/s. Ankle and knee joint kinematics were calculated using cardan angles. Kinematic variability of the ankle and knee joints was evaluated using traditional methods (standard deviation and coefficient of variation) and a non-traditional method (spanning sets). **Results**: There was a significant difference in transverse plane knee kinematics as an interaction effect of gender and condition ($F = 4.544$, $P = .043$). There were no significant differences upon other kinematic parameters of the ankle or knee data as an interaction effect between gender and condition. There was a significant difference in transverse plane knee kinematic variability measured via spanning set as an interaction effect of gender and condition ($F = 5.306$, $P = .029$). There were no other significant differences in kinematic variability measures of the ankle or knee data as an interaction effect between gender and condition. **Conclusion**: It is not clear if one could clearly suggest or refute the use of an OTC orthotic for direct control of ankle and knee mechanics. For some the less expensive OTC device may provide an inexpensive alternative to a higher priced custom device. For others there may be a definite need for a more custom fit appliance. There is no reason to suspect the changes in kinematics and variability reported in this study are the effect of a pathologic (abnormal) control strategy. It is not completely clear as to whether or not the SS method used is more sensitive to movement variability.

Acknowledgements

This has process has been long, humbling, rewarding, discouraging, exciting, rousing, uninspiring, taming, gratifying, disappointing, overwhelming, and awesome. This process has been all things emotionally. It has at times consumed all of my attention and at times been least deserving. To say the least, this project has taught me more about myself and forced me grow in ways I never expected. For that, I have to acknowledge the process itself. I respect it now.

I want to thank my wife (Katie) and daughter (Kayla) for showing me what is important and keeping me focused along the way. The last few years have been difficult and have inspired personal growth in each of us. This time, this life, has opened my eyes to possibilities I never knew existed. Along the way there has been a lot of physical and mental pain and I could not have endured without either of you.

To the committee, thank you for remaining fair and balanced. There were times along the way when I felt abandoned, alone, or lost. You never turned me away, regardless of personal convictions or reservations. I am not the easiest student to have. I may not share your vision but I do appreciate your passion. I only wish I had found what I was passionate about earlier. I guess these things take time.

I don’t really drink. I don’t smoke. So I can’t thank the Jim’s and Jack’s or the Marlborough men of the world that help one drown sorrow and self-loathing. I gave up on that. I now feel connected to myself, my search for knowledge, and my connection to people. To everyone involved in making adult me, me…Thank You.

Definitions

Abduction: A motion that pulls a structure or part away from the midline of the body.

Acceleration: The rate of change of velocity with respect to time.

Adduction: A motion that pulls a structure or part towards the midline of the body.

Angular motion: Motion that is not linear. If the axis of rotation is fixed, all particles in the body travel in a circular manner. If the axis of rotation is not fixed the motion is actually a combination of translation and rotation

Angular Velocity: The rate of movement in rotation calculated as the first time derivative of angular displacement.

Displacement: The change in the position of a body.

Dorsiflexion: Extension of the entire foot superiorly.

Eversion: The movement of the sole of the foot away from the midline of the body.

Extension: A straightening movement that increases the angle between body parts.

External Rotation: Lateral rotation or rotation away from the center of an object or body.

Flexion: Bending movement that decreases the angle between two parts.

Force: A vector quantity that describes the action of one body on another.

Frontal Plane: Divides the body into back and front portions.

Gait: A particular way of moving on foot.

Ground Reaction Force: A force that acts upon the body as a result of interaction with the ground.

Internal Rotation: Medial rotation or rotation towards center of an object or body.

Inversion: the movement of the sole of the foot towards the midline of the body.

Joint Forces: The forces that exist between the articular surfaces of the joint. Joint forces are the result of muscle forces, gravity, and inertial forces.

Joint Reaction Forces: The equal and opposite forces that exist between adjacent bones at a joint caused by the weight and inertial forces of the two segments.

Kinematics: The description of motion.

Kinetics: The study of the forces that cause motion

Lever: A lever is a system that tends to change the mechanical advantage of an applied force.

Linear motion: When all parts of the body travel along parallel paths.

Locomotion: The act of moving from place to place.

Pronation: Of the foot, a combination of eversion, abduction, and dorsiflexion.

Plantarflexion: Flexion of the entire foot inferiorly.

Reference Frame: An origin and a set of coordinate axes that serve as the basis for the calculation of displacement and its derivatives.

Rotation: A motion that occurs when a part turns on its axis.

Rigid body: A collection of particles occupying fixed locations with respect to each other.

Sagittal Plane: Divides the body into left and right portions.

Supination: Of the foot, a combination of inversion, adduction, and plantarflexion.

Transverse Plane: Divides the body into head and tail portions.

Vector: A quantity or force that has both direction and magnitude.

Velocity: A measure of a body's motion in a given direction. Because velocity has both magnitude and direction, it is a vector quantity that can be positive, negative, or zero.

Chapter One

1.1 Introduction

Many people engage in daily physical activity in an attempt to live a healthy life. Jogging and running are popular modes of exercise. This popularity is coupled with overuse injuries of the foot, lower-leg, knee, hip and even low back, associated with the high levels of stress placed on the lower extremity through jogging and running (Gross, 1998; Rodgers, 1988; VanBoeren and Sangoerzan, 2003; Vaughan, 1984).

Research studies have been conducted in an attempt to better understand the mechanisms of such chronic injuries. The structure and function of the lower extremity (the segments of the foot, ankle, shank, knee, and thigh and the joints of the ankle, knee, and hip) have been examined using static (clinical postural assessment) and dynamic (gait analysis) methods in an attempt to identify predispositions for overuse injury. Much effort has been focused on exploring the anatomical structure of the foot and its functional role during locomotion.

The foot has an amazingly intricate and complex structure for its function during locomotion. The foot and ankle make up a complex anatomical structure consisting of 26 irregularly shaped bones, 30 synovial joints, more than 200 ligaments, and 30 muscles acting on the segments (Hamil and Knutzen, 2009). The foot can be sectioned into three compartments: the hindfoot (calcaneus and talus), the midfoot (navicular, talus, three cuneiform bones, five metatarsals), and the forefoot (the proximal, middle, and distal phalanges). Within the bony and soft tissue structure are the medial, lateral, and longitudinal arches, all of which are believed to play a major role in the primary function of the foot - which is to transmit loads between the lower leg and ground (Nuber, 1989). Dynamically, such as during human locomotion, the foot acts as both a shock absorber and a mechanical propeller. When any of the anatomical structures of the foot and ankle act improperly, humans may be more likely to sustain overuse type injuries throughout the lower extremity.

While the foot plays an important role in locomotion, the movement of the ankle joint complex has a more significant influence on gait mechanics: movement deficiencies at the ankle joint often contribute to abnormal gait kinematics. The basic movements about the ankle joint are plantar flexion /dorsiflexion (which occur in the sagittal plane about the talocrural joint) and pronation /supination (which occur in all three planes of motion about the subtalar joint). Pronation is produced by eversion, abduction, and dorsiflexion, while supination is produced by inversion, adduction, and plantarflexion; these motions occur in the frontal, transverse, and sagittal planes, respectively.

Rearfoot motion is thought to be coupled with movements of other joints of the lower extremity. Previous studies have indicated a coupling effect between rearfoot inversion and eversion with external and internal rotations of the tibia, respectively (McClay and Manal, 1995; Munderman et al., 2003; Nawoczenksi et al., 1995; Nigg et al., 1998). Because of this relationship, pathologic rearfoot movement patterns often lead to overuse injuries of the lower extremity such as plantar fasciitis, Achilles tendonitis, patellofemoral pain, and stress fractures. One intervention used to correct deficient foot and ankle kinematics are foot orthoses.

Foot orthoses, or shoe inserts, are designed to help restore “normal” function of the foot and ankle by providing support to the affected arch(es) and bony structures. These orthoses or inserts are more commonly referred to as orthotics. There are numerous types/styles of orthotics available. Orthotics can be soft, allowing for some natural movement of the foot, or rigid, restricting natural movement and placing the foot in a "proper" functional alignment. Both soft and rigid orthotics can be custom made, which requires the consultation of a specialist (such as a podiatrist, physical therapist, athletic trainer, or orthotist) and are often expensive to fabricate. Soft orthotics are commonly used for shock absorption/attenuation and minor cases of motion control. Rigid orthotics are designed primarily for motion control, with less consideration given for comfort; this results in a more durable, but more expensive orthotic device (Razeghi and Batt, 2000) . Clinically, the use of orthotic devices is widely accepted, while the mechanism for clinical effectiveness is not yet fully understood.

Recent literature has been published acknowledging the contrasting results, and have begun to examine how these results may be related to the inherent variability of human movement, necessary to maintain musculoskeletal health (Ferber et al, 2005; Kurtz and Stergiou, 2001; Harbourne and Stergiou, 2009.).

1.2 Problem Statement

The purpose of this study was to investigate the gender-related differences and lower extremity kinematic variability when using an over-the-counter (OTC) soft orthotic device during treadmill running in male and female recreational runners with pes planus (flat feet). Subjects were required to run on a treadmill at a speed of 3.35 m/s; lower extremity kinematic data was recorded using an active opto-electronic motion capture system. Traditional (standard deviation and coefficient of variation) and a non-traditional (the spanning set) measures of variability were calculated for all kinematic variables of interest (angular displacement [range of motion] and angular velocity in the sagittal, transverse, and frontal planes for both the ankle and knee joints).

The specific purposes of this study were to:

(1) determine the effects of footwear condition (orthotic versus non orthotic) on lower extremity kinematics during treadmill running in male and female recreational runners,

(2) determine the effects of gender (male versus female) on lower extremity kinematics during treadmill running,

(3) determine the effects of footwear (orthotic versus non orthotic) and gender (male versus female) on lower extremity kinematic variability during treadmill running,

(4) compare traditional measures (standard deviation and coefficient of variation) and non-traditional measures (the spanning set) of lower extremity kinematics variability during treadmill running in male and female recreational runners.

1.3 Variables

The independent variables in this experiment were:

(1) Orhotic condition: each subject performed one running trial without the orthotic and one running trial with the orthotic.

(2) Gender: 15 male and 15 female recreational runners participated in this study to determine the existence of any gender-related effects.

The dependent variables in this experiment were:

(1) Angular displacement [range of motion] and angular velocity in the sagittal, transverse, and frontal planes for both the ankle and knee joints and,

(2) Variability measures (standard deviation, coefficient of variation, spanning set) of the angular displacement (range of motion) data in the sagittal, transverse, and frontal planes for both the ankle and knee joints.

1.4 Hypothesis

It was hypothesized:

(1) There will be no difference in lower extremity kinematics (ankle and knee joint ROMs) as a function of footwear condition (orthotic versus no orthotic).

(2) There will be no differences in lower extremity kinematics (ankle and knee joint ROMs) as a function of gender (male versus female).

(3) There will be no difference in lower extremity kinematic variability as a function of footwear condition (orthotic versus no orthotic) and gender (male versus female).

1.5 Assumptions and limitations

The assumptions made in this study were:

(1) All runners were healthy and free of musculoskeletal injury.

(2) All participants used a rearfoot-strike running style.

(3) Randomness of order of foot wear condition would have no effect on kinematic/proprioceptive response while running.

The limitations to this study were;

(1) No kinetic analysis was conducted in conjunction with kinematics data.

(2) The subjects did not have a long-term program to adapt to the orthotics (only a 15 minute running trial).

(3) Only subjects with clinically identified flat feet participated in this study.

Chapter Two

It is important to understand how the structure of the foot relates to the function and biomechanics of the lower extremity. This concept of anatomical and biomechanical connection is often referred to as the kinetic chain. Since the foot serves as the link between the ground surface and the body, the foot is believed to have a significant effect on lower extremity function including; adaptation to terrain, proprioception, and leverage for propulsion (Dugan and Bhat, 2005). It is accepted that the anatomy of the foot and ankle have an influence on the biomechanics of the knee and hip. In general, the lower extremity is comprised of 30 bones of the foot, lower leg, and thigh.

2.1 Anatomy of the lower extremity

2.1.1 Anatomy of the foot and ankle

The foot consists of 26 bones – tarsals, metatarsals, and phalanges - and can be divided into three sections: rearfoot (hindfoot), midfoot, and forefoot.

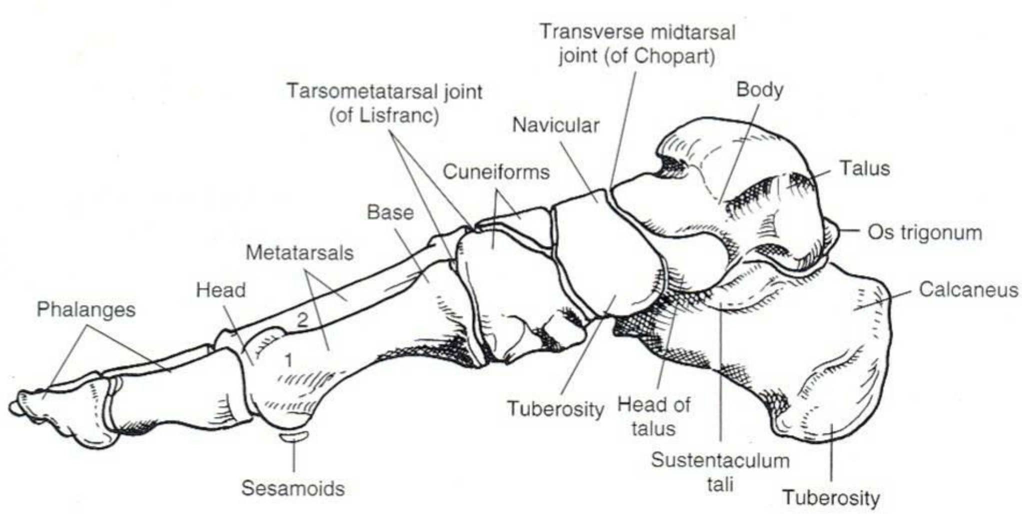

Figure 2.1 Bony anatomy of the foot, medial aspect (from Sammarco and Hockenbury, 1989)

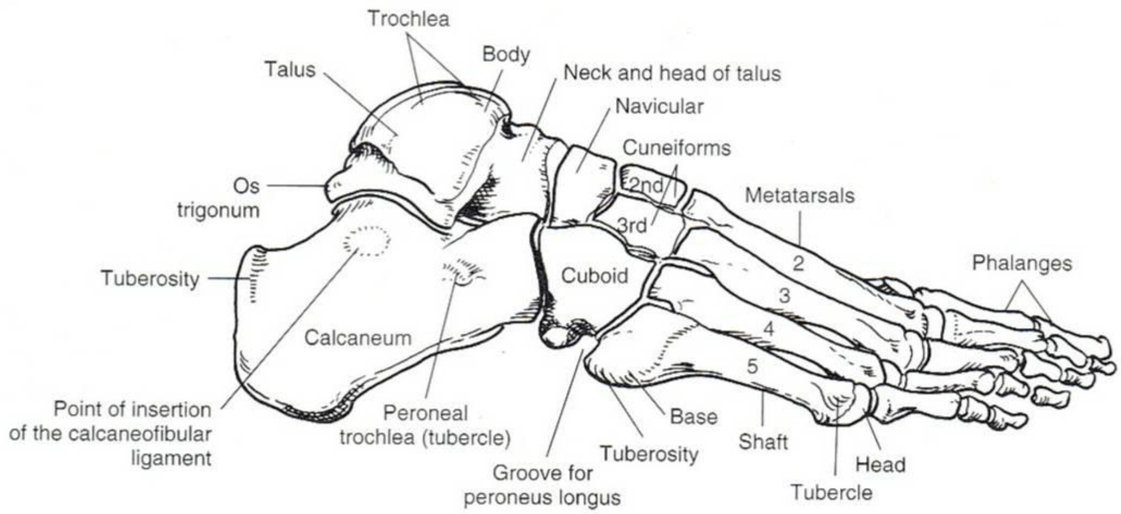

Figure 2.2 Bony Anatomy of the foot, lateral aspect (from Sammarco and Hockenbury, 1989)

Functionally the foot contains three arches: medial longitudinal arch, lateral longitudinal arch, and transverse metatarsal arch. These arch structures offer the foot flexibility and adaptability to dynamic activity, such as gait. The kinetic link between the foot and lower leg is typically referred to as the ankle joint or talocrural joint where most of the available motion occurs. During gait the movements about the subtalar and midtarsal joint in conjunction with the talocrural joint contribute to the overall movement of the foot and ankle that translate throughout the lower extremity.

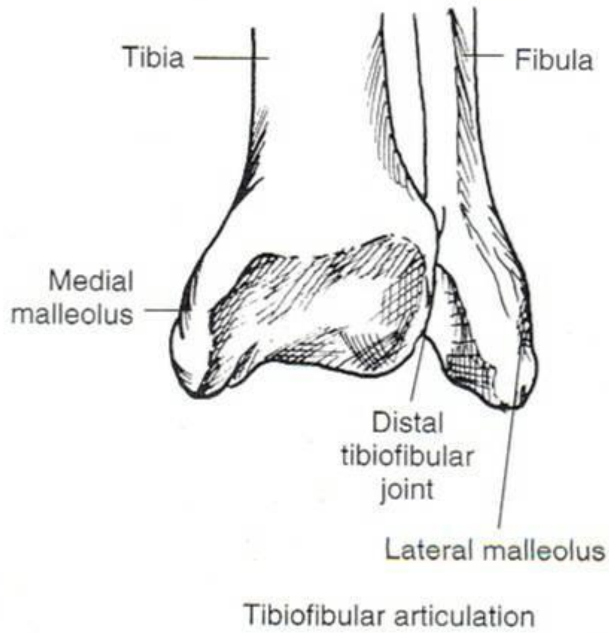

Figure 2.3 Bones of the lower extremity (tibia and fibula) that connect to the talus of the foot to make up the talocrural joint (From Sammarco and Hockenbury, 1989)

The rearfoot segment of the foot is where numerous studies have attempted to correlate the relationship between excessive joint motion with running injuries (Kurz and Stergiou, 2004; Nawoczenksi et al., 1995; Nigg et al., 1998; Perry, 1983; Smith et al., 1986). The rearfoot is formed by two tarsal bones, the calcaneus and the talus, and is thought to provide stability and shock absorption during the initial stance phase of heel strike of gait. Motion that occurs in the rearfoot is due to movement of the calcaneus about the talus occurring in the frontal plane and is known as calcaneal inversion or eversion. It is thought that abnormal or excessive calcaneal motion of the rearfoot leads to altered mechanics of the foot and its structures which in turn translates proximally throughout the kinetic chain of the lower extremity (Perry, 1983).

The midfoot segment is composed of the navicular, cuboid, and the lateral, intermediate, and medial cuneiforms; which are all considered to be tarsal bones. The midfoot segment is highly flexible in dynamic situations and any altered function of this structure primarily affects the height of the medial longitudinal arch. Lower than normal medial longitudinal arches result in a condition known as pes planus, or flat foot, deformity; individuals with this condition are at higher than normal risk of injury due to the excessive amounts of pronation that occur through the midstance phase of running gait (Cowan et al., 1993).

The forefoot segment is formed by the five-metatarsal bones along with the 14 phalanges. The primary role of the forefoot is to act as a lever for propulsion during the pre-swing phase of gait (Sammarco and Hokenbury, 1989). Movement that occurs within the forefoot segment, just as in the midfoot and rearfoot segments, is not isolated from other segments of the foot. Foot motion that is segment specific is linked to movements of the other foot segments, making foot movement extremely dynamic, especially in gait patterns.

All 26 bones of the foot are bound together through ligamentous support so that none of the bones within the foot translate weight forces directly against the ground surface in dynamic motions and instead translate forces upon each other and throughout the lower extremity. The shape of the bones, along with the ligamentous and muscular support system, form the medial and lateral longitudinal arches and the less pronounced transverse arch (Chan and Rudins, 1994). The arches of the foot allow for force distribution in both static and dynamic conditions by allowing the foot to become more or less rigid as necessary for load acceptance and transfer.

In non-weight bearing conditions, the arches of the foot are more noticeable than in weight bearing conditions. The arches will flatten as the foot supports the weight of the body during locomotion. During weight-bearing activity (locomotion), pronation occurs throughout the medial longitudinal arch and is often the focus of lower extremity and foot pathology resulting from excessive motion. The medial longitudinal arch is formed by the calcaneus, talus, navicular, medial cuneiform, and the first metatarsal. In addition to the bony structure, ligamentous and muscular support allow for greater amounts of motion than the other arches. The lateral longitudinal arch is formed by the calcaneus, cuboid, and the fifth metatarsal and is rarely associated with injury. The transverse metatarsal arch is formed by the lengths of the metatarsals and tarsals assuming its concave shape from anatomical structures -the heads of each metatarsal to the calcaneus.

The ankle joint is the functional link between the foot and the lower leg. The ankle joint is the articulation of the bones of the lower leg - the tibia and the fibula, along with talus. These bones also make up what is known as the talocrural joint (see figure 2.4). In addition the ankle consists of another joint, the subtalar joint. Due to the composition of bony, ligamentous, and muscle structure of the foot and lower leg, movements about each are thought to be influential on the other. Therefore it is unrealistic in dynamic situations to discuss foot or ankle motion without consideration of each other.

2.1.2 Anatomy of the lower leg and thigh

The lower leg is made up of tibia and fibula. The lower leg is connected to the thigh by the knee joint which is an articulation between the proximal head of the tibia and the distal end of the femur. Two menisci lie between the femur and the tibia to enhance the contact surfaces of each bone to provide cushioning, absorb shock in weight bearing conditions, and allow for fluid joint movements. Another bone of importance of the knee joint is the patella, a sesamoid bone located within the quadriceps tendon near its attachment on the tibia. Runners are often at risk of patello-femoral pain (PFP) syndrome, a painful condition thought to be caused due to contact of the posterior surface of the patella along the femur as a result of. an increased Q-angle or excessive pronation (Eng and Pierrynowski, 1994).

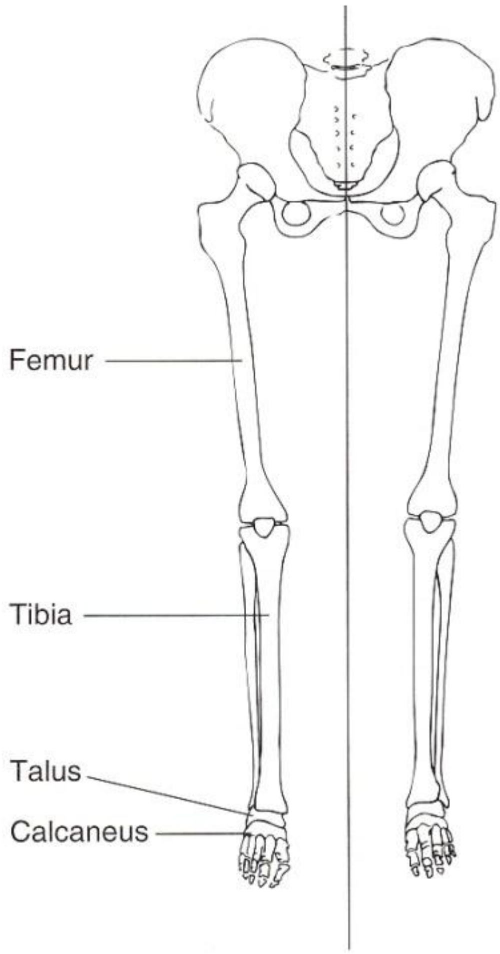

Figure 2.4 Bony anatomy of the lower extremity including pelvic girdle (Sammarco and Hockenbury, 1989).

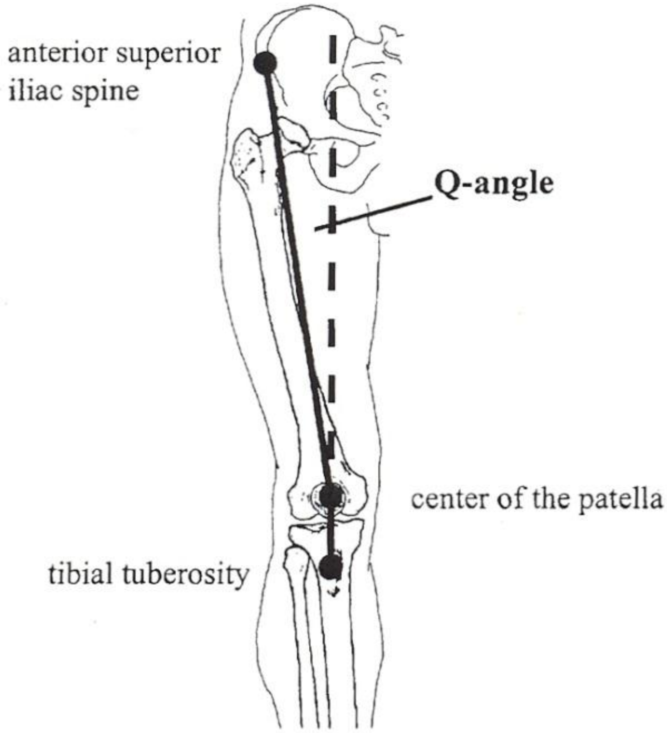

Figure 2.5 Depiction of Q-Angle measurement from the anterior superior iliac spine to the tibial tuberosity (from Chans and Rudins, 1994).

The femur is connected to the appendicular skeleton through the hip joint, where the femur meets the pelvis. The pelvis is composed of the illium, ischium, and pubic bones. The proximal head of the femur is connected to the pelvis via a concave attachment site on the illium, called the acetabulum.

2.2 Foot structure and function

Integrity and mobility of the foot are essential for all forms of gait. Faulty foot mechanics from subtalar and midtarsal joint dysfunction may have significant effects on the overall function of the lower extremity throughout the body (Hreljac et al., 2000). Lang, Volpe and Wernicke (1997) reported that the subtalar and midtarsal joints are often the primary sites of the lower extremity kinematic chain for compensation of faulty foot mechanics. It was noted that the subtalar joint provides a structural and mechanical link between the ankle (talocrural joint) and the midtarsal joints, performing the function of converting lower limb (shank) rotation into pronation and supination. It is likely that since the foot provides a link between the ground and the lower extremity during gait structural abnormalities can contribute to the onset of injuries of not only the foot and lower leg but of the knee, hip, and back as well.

Foot structure and function is of primary concern during the stance phase of running; faulty structure and function predisposes individuals to running- related injuries. It has been reported that more activity-related injuries are related to flat foot or pronated (pes planus) foot structures (Subotnick, 1981). However, other studies have reported that a high arched or everted (pes cavus) foot structure creates a less flexible foot leading to mechanical deficiencies of the lower extremity and ultimately injury. Therefore, it seems that there is much to be understood about the specific effect of arch height and running injuries.

Nigg, Cole, and Nachbauer (1993) found no correlation between maximal rear foot eversion and arch height (pronation) yet did report a significant correlation between maximal internal leg rotation and arch height. A significant correlation was found between the transfer of foot eversion to internal leg rotation with increasing arch height. The primary finding of this study is that foot eversion and internal rotation of the tibia are coupled during running. This suggests that although arch height may not be directly related to running injures, the effect of arch height on foot eversion may be a causal factor in knee -elated pain and other injuries during running.

2.3 Kinematics of the foot and ankle

Movement that occurs about the ankle or talocrural joint during running occurs in all three cardinal planes, giving it three degrees of freedom. Dorsiflexion and plantarflexion occur in the sagittal plane, inversion and eversion of the rear foot occur in the frontal plane, and internal rotation (also known as forefoot adduction) and external rotation (also referred to as forefoot abduction) occur in the transverse plane. Since the axes of the foot and ankle are not perpendicular to any of the cardinal planes, all motion is considered tri-planar and in some cases uniaxial.

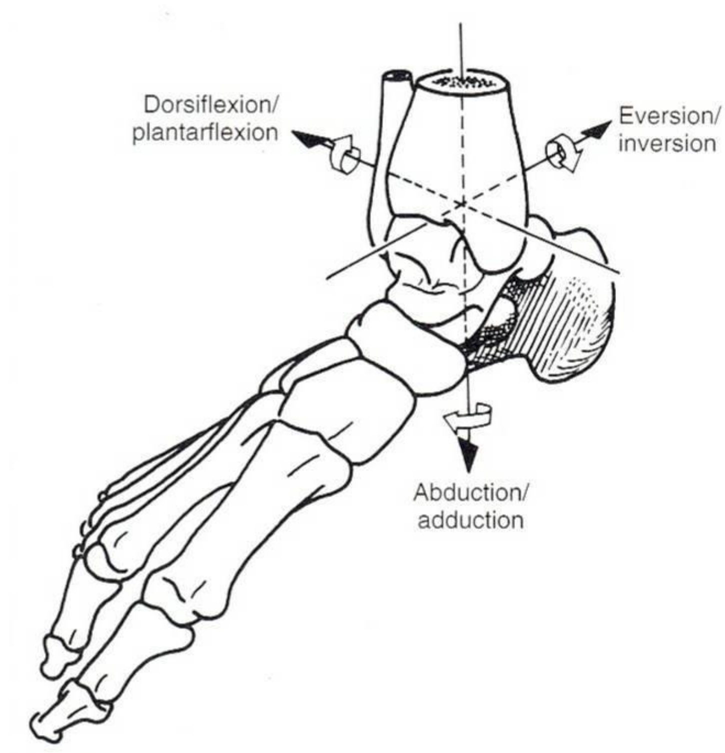

Figure 2.6 Foot motion available around three axes (Sammarco and Hockenbury, 1989).

The tri-planar motions about the foot and ankle are pronation, the result of eversion, abduction, and dorsiflexion, and supination, the result of inversion, adduction, and plantarflexion (Chan and Rudins, 1994).

During gait, when the foot comes into contact with the ground it is “unlocked” giving it the ability to adapt to terrain as necessary. Later in the stance phase, when the foot is preparing for push-off, it becomes less flexible creating a functional lever to propel the body forward. Rotations about the lower extremity, particularly external rotations, are thought to contribute to the overall rigidity of the structure of the foot. Dorsiflexion of the ankle while the lower extremity is in a closed kinetic-chain activity causes internal rotation of the lower leg (tibia) and pronation of the foot (Chan and Rudins, 1994). External rotation of the leg produces inversion about the calcaneus, and internal rotation of the leg causes eversion of the calcaneus.

During gait, the bones of the rear foot are normally in line with the leg and perpendicular to the ground. Higher values of dorsiflexion are associated with higher gait velocities. Throughout the stance phase, the foot pronates to accommodate the transfer of forces from the body to the ground (Dugan and Bhat, 2005). As the foot pronates, the leg internally rotates about the foot and ankle. Likewise in opposite motion patterns, as the foot supinates, the leg must externally rotate about the foot and ankle. During gait, rotations about the lower extremity occur in the transverse plane. Each rotation motion progressively increases from the more proximal segments to the more distal segments: rotation about the tibia will be greater than femoral rotation (Chan and Rudins, 1994).

During running, pronation of the foot is completed about five times faster than in walking, occurring usually within the first 30 ms of the stance phase. Actual values may vary depending on running speed (Dugan and Bhat, 2005). A basic kinematic description of the foot and lower leg during running is as follows;

1) at heel strike, the calcaneus is inverted and the lower leg is internally rotated,

2) as the foot begins to absorb the load of the body, the calcaneus everts resulting in pronation of the foot and allowing the foot to become more flexible,

3) after maximum pronation, the foot begins to supinate and externally rotate the lower extremity,

4) external rotation of the lower extremity causes inversion of the calcaneus creating a rigid lever type surface (Chan and Rudins, 1994).

This pronation mechanism of the foot is vital for the absorption and transfer of forces throughout the lower extremity. High arched, or pes cavus, foot structures have less flexibility, and less range of pronation motion than a normal foot structure, causing this type of foot to be less effective at force absorption. This likely contributes to why individuals with high arched, foot structures are believed to be susceptible to injuries while running.

2.4 Ankle kinematics and excess pronation

McClay and Manal (1998) presented kinematic data showing there to be a significant difference in the position of the rearfoot at heelstrike and at toe off. Even thought the total range of eversion was not significantly different compared to the normal group. This data presents us with a unique view of the kinematics difference of excess pronation.

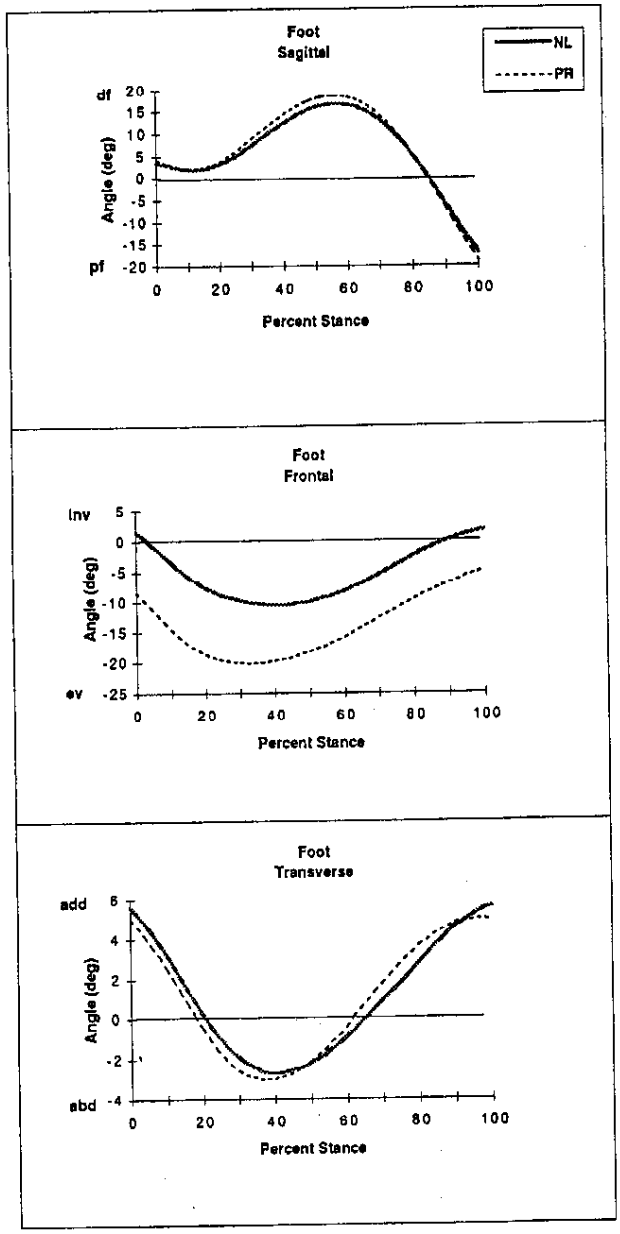

Figure 2.7 A comparison of three-dimensional kinematics data comparing ranges-of-motion between excessive pronators (dashed line) and normal runners (solid line) about the foot/ankle. Each range of motion presented along the vertical axes and stance phases normalized as a percentage of time is presented along horizontal axes. (from McClay and Manal, 1998).

2.5 Kinematics of the knee

Kinematics of the knee are less complex compared to those of the foot and ankle, but no less important when looking at the functional integrity of the lower extremity during gait. Movement that occurs about the knee joint during running gait occurs in all three cardinal planes, giving it three degrees of freedom. Flexion and extension occur in the sagittal plane, abduction (genu valgum) and adduction (genu varum) occur in the frontal plane, and internal rotation and external rotation occur in the transverse plane. All motions of the knee joint are motions that occur between the tibia and femur in relation to each other.

The primary motion of the knee during gait occurs in the sagittal plane as the knee flexes and extends. The total range of motion required for effective gait is dependent upon the speed of movement. The findings of Novacheck in 1998 seem to support other another report of kinematic change as running velocity increases. The flexion angle of both the hip and knee were shown to demonstrate an increase in each respective flexion angle (Pink et. al., 1994).

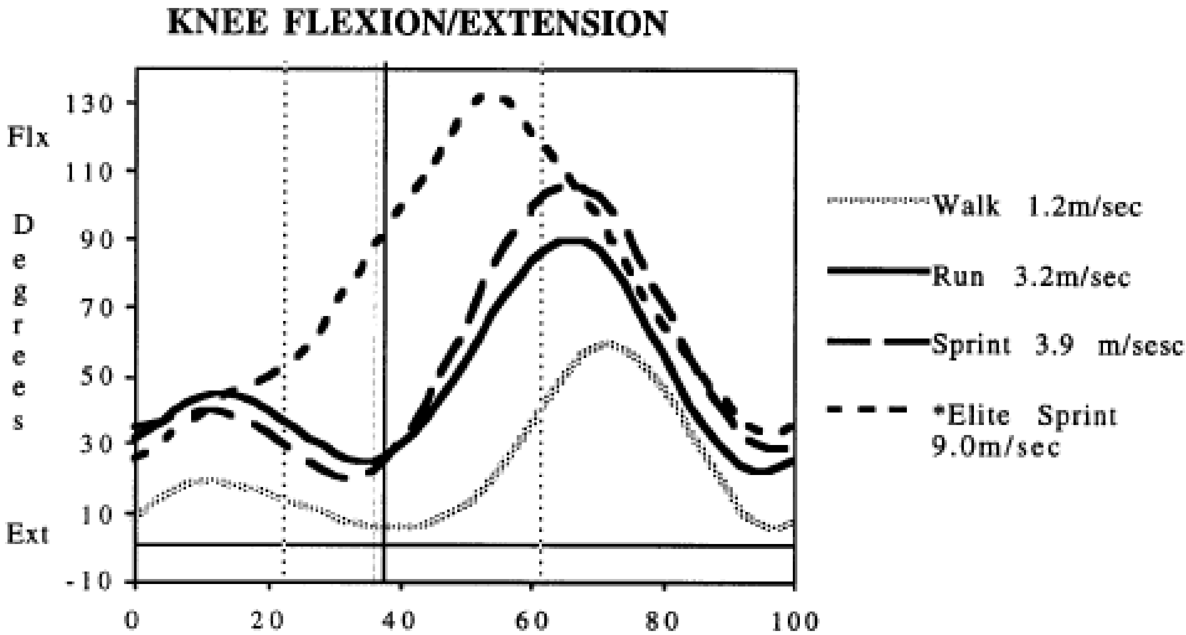

Figure 2.8 Comparison of sagittal plane knee kinematics dependent upon gait velocity normalized to a percentage of stance phase (from Novachek, 1998).

2.6 Knee kinematics and excess pronation

The study in 1998 from McClay and Manal also provided a detailed look at the different kinematic profiles about the knee in normal and runners classified as excessive pronators. For our study the discrete measures observed do not provide much context outside of the overall kinematic profiles. The authors (McClay and Manal, 1998) do mention a positive correlation between rearfoot eversion and knee internal rotation excursion. This provides some insight into the coupling effect of the ankle joint about the knee.

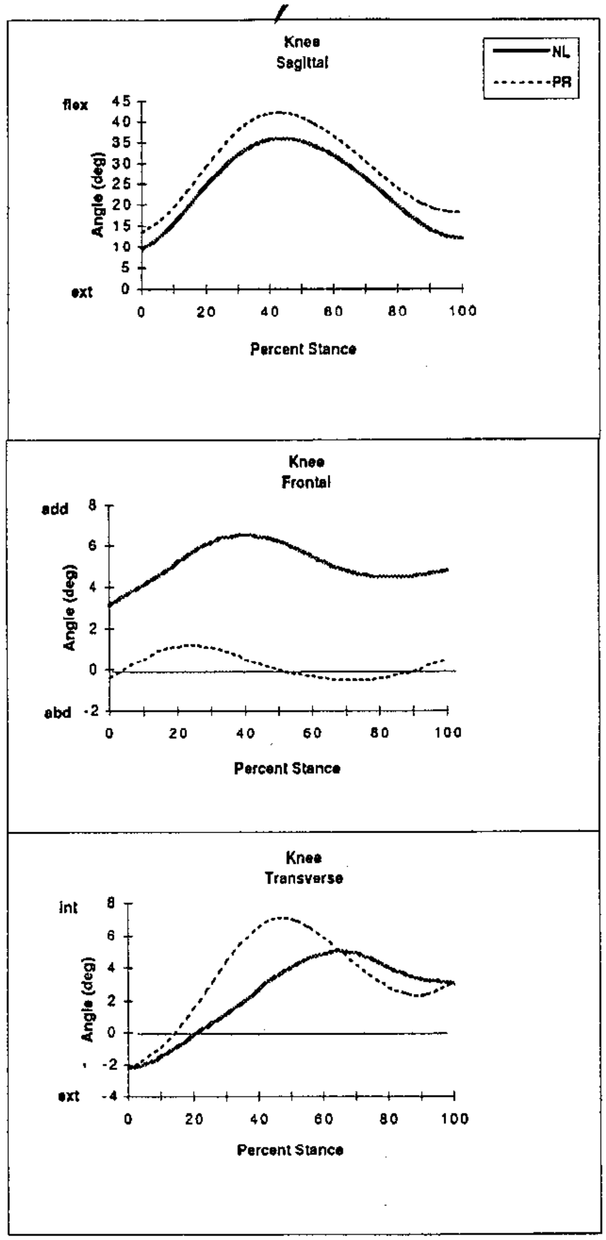

Figure 2.9 A comparison of three-dimensional kinematics data comparing ranges-of-motion between excessive pronators (dashed line) and normal runners (solid line) about the knee. Each range of motion presented along the vertical axes and stance phases normalized as a percentage of time is presented along horizontal axes. (from McClay and Manal, 1998).

2.7 Orthotic devices

An orthotic is described as a type of "shim" or wedge placed between the foot and show to modify foot position during the support phase of running (Bates etal., 1979). The material used for these devices can either be made from flexible materials to more rigid plastics. Foot orthotic devices are prescribed as a modality to alter lower extremity biomechanics during gait. There are three categories of orthotics: hard, semi-rigid, and soft. Each of these categories are intended to provide a specific function. Hard orthotics are designed primarily for motion control, semi-rigid orthotics are designed to control motion and provide some cushioning, and soft orthotics are designed less for motion control and more for cushioning. Both semi-rigid and soft orthotics are commonly used for physical activity and sport applications (Smith et. al., 1986).

The efficacy of orthotic prescription in the treatment of injuries associated with running and foot structure has been accepted in the medical community as a valid treatment option. However, it is still unclear exactly how foot orthotics effect lower extremity mechanics as an intervention technique to reduce the risk or occurrence of chronic injuries (Kilmartin and Wallace, 1994; MacClean, 2001; VanMechlen, 1992).

Numerous studies have been conducted examining the influence of orthotic devices on lower extremity kinematics during running (Bates et al., 1979; Brown etal., 1995; Eng and Pierrynowski, 1994; Ferber et al., 2005; Munderman et al., 2003; Nawoczenski et al., 1995; Nigg et al. 1997; Stackhouse et al., 2004; Stacoff et al., 2000; Williams et al., 2003;). Early studies(Bates et al., 1979; Brown et al., 1995) focus on the efficacy of an orthotic device to control pronation in runners. More recent studies (Eng and Pierrynowski, 1994; Ferber etal., 2005; Munderman etal., 2003; Nawoczenski etal., 1995; Nigg et. al. 1997; Stackhouse etal., 2004; Williams etal., 2003) have taken into account the effect of an orthotic on not just rearfoot motion but on the coupling of rearfoot motion (ankle) with tibial rotations (knee)

2.8 Effect of orthotics on ankle kinematics

Stackhouse, McClay-Davis, and Hamill (2004) used three-dimensional kinematic techniques to examine the differences between forefoot and rear foot strike patterns in runners with and without foot orthotic devices. The joint specific kinematics from this study was used for comparison. The orthotic device used in this study is reported as being a semi-rigid device, which may not have the exact effect of a soft orthotic device but may still allow for a similar comparison between our shod and orthotic running conditions. Only the kinematic data presented in this study regarding rear foot strike patterns is used for reference. The kinematics reported in this report are consistent with the kinematics reported by McClay and Manal (1998) where kinematics were compared between groups classified as normal and excessive pronators.

In the sagittal plane, the ankle of a rearfoot striker at heel strike is in dorsiflexion. The actual angle of dorsiflexion at heel strike tends to increase as the speed of ambulation increases (Nawoczenski et al., 1995). This finding runs counter to that of a comparison of “slow” and “fast” self-selected running paces (defined as either above or below a pace of 3.35 m/s) where no difference in sagittal plane ankle kinematics were noted (Pink et al., 1994).

As the stance phase progresses, the ankle plantarflexes until it reaches close to a neutral position at around 20-25 percent of the stance phase or during the support phase. The actual time the support phase is reached can vary between runner and between running speeds where shorter support phases are seen in higher running speeds (Pink et al., 1994). The ankle then prepares the foot for propulsion as it nears terminal stance and toe-off through dorsiflexion. As the stance phases nears its end, the ankle plantar flexes through toe-off. Once the foot has left the ground the stance phase is complete. The approximate values of the ranges of motion of the ankle in this plane are, near 20 ° of dorsiflexion at heel strike and near 10° of plantarflexion at toe off (Stackhouse et al., 2004; Pink et al., 1994).

The use of an orthotic device had little effect on the sagittal plane motion at the ankle joint with respect to the overall range of motion throughout the stance phase. It was noted that at heel-strike the average values of dorsiflexion were slightly lower (three degrees) with the orthotic device compared to without the orthotic device. Likewise the plantarflexion values at toe-off were slightly greater (one or two degrees) when using the orthotic device (Mundermann et al., 2003).

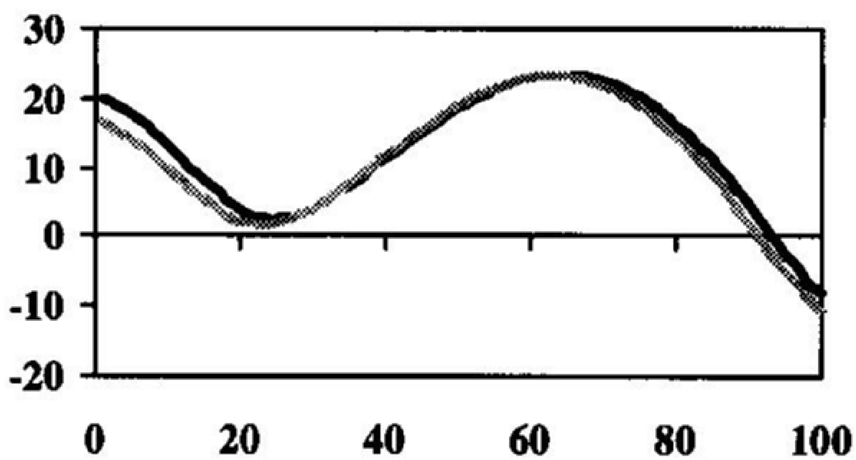

Figure 2.10 Graphical representation of ankle joint kinematics in the sagittal plane as the ankle begins in plantarflexion. Presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et al.,2004).

In the frontal plane, the ankle joint is inverted near heel strike. As the foot progresses through the stance phase the ankle undergoes eversion (Chan and Rudins, 1994). Since the true ankle joint (talocrural) is primarily responsible for movements occurring in the sagittal plane, this frontal plane motion is believed to occur at the subtalar joint of the ankle. Near maximal eversion occurs close to 50 percent of the total duration of the stance phase during running. The approximate range-of-motion values reported are around 8 degrees of inversion at heel-strike and close to 10 degrees of eversion just after the loading phase as the foot is preparing for propulsion. It is reported that with the use of an orthotic device, the maximum amount of calcaneal eversion is reduced up through mid-stance, yet there appears to be little effect on the timing of the peak eversion when using an orthotic device (Stackhouse et al., 2004).

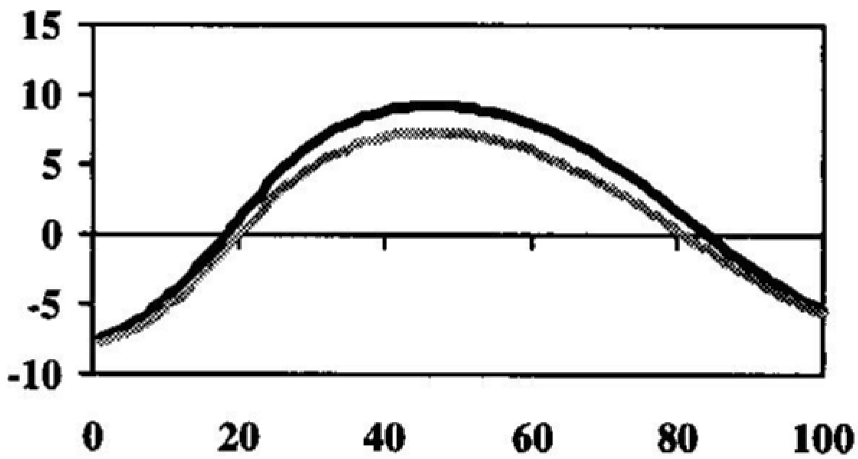

Figure 2.11 Graphical representation of ankle joint kinematics in the frontal plane as the ankle begins in an inverted position and progressively everts. Range of motion is presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et al., 2004).

In the transverse plane the ankle joint is close to its neutral position, however it is slightly abducted near heel strike. As the foot progresses through the stance phase the ankle abducts even more up through the support phase. As the phase progresses, the ankle abducts even further until the ankle and foot prepare for toe-off, at which point the ankle begins to adduct until the stance phase is complete. Near maximal abduction during running occurs near 50 percent of the total time duration of the stance phase (Nigg et al., 2003). The approximate range-of-motion values reported are around 1 degree of abduction at heel-strike and close to 5 degrees of abduction just after the loading phase as the foot is preparing for propulsion. It is reported that with the use of an orthotic device, the maximum amount of ankle abduction is increased through mid-stance, and may affect the timing of the peak abduction when using an orthotic device (McClay and Manal, 1998; Stacoff et al., 1999, Stackhouse et al., 2004).

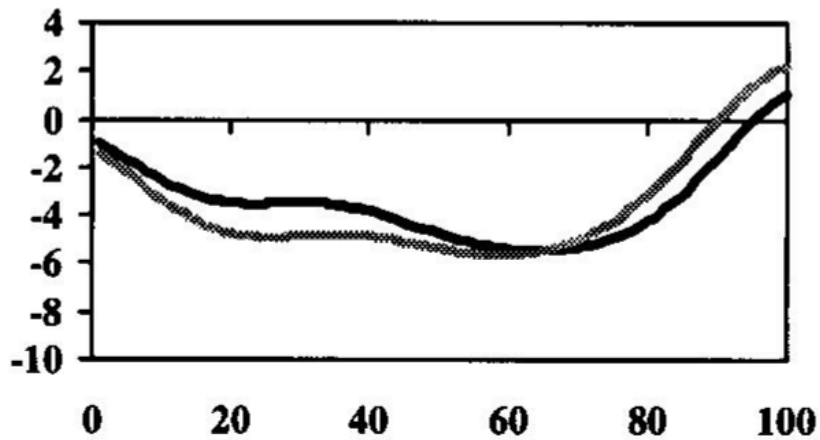

Figure 2.12 Graphical representation of ankle joint kinematics in the transverse plane as the ankle begins slightly abducted and continues to abduct until near toe-off. Range of motion is presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et. al., 2004).

2.9 Effect of orthotics on knee kinematics

In the sagittal plane, the knee flexes and extends. At heel-strike the knee is in slight flexion, yet is usually reported as being near maximal extension. The knee continues to flex as the stance phase progresses reaching its maximum flexion angle between 40 and 50 percent of the total duration of the stance phase (Nigg et al., 2003). As the lower extremity is preparing for propulsion and toe-off, the knee joint again extends, and at toe-off the joint angle is approximately the same as at heel-strike. The ranges reported for this study were approximately 10 degrees of knee flexion at the time of ground contact with a maximum flexion angle close to 45 degrees.

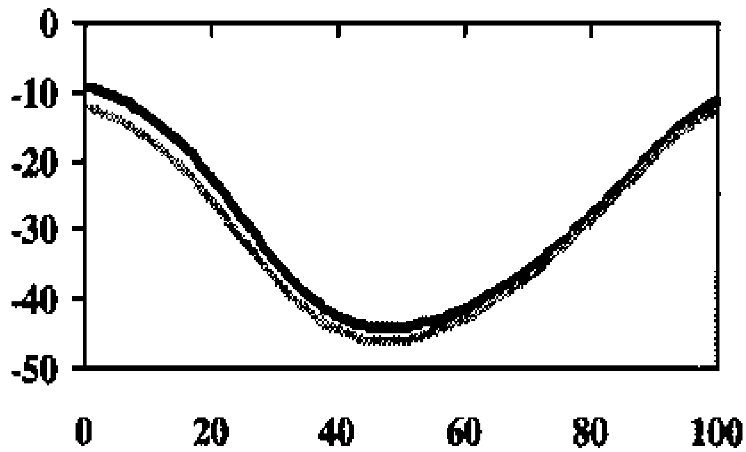

Figure 2.13 Graphical representation of knee joint kinematics in the sagittal plane as the knee goes through its flexion range of motion. Range of motion is presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et al., 2004).

In the frontal plane, the knee abducts and adducts. At heel-strike the knee is in a state of abduction and progressively adducts, or brings the tibia closer towards the midline of the body with respect to the femur, as the stance phase progresses. In the frontal plane, there appear to be two peaks in the adduction motion of the knee joint: the first occurs just after 30 percent of the total duration of stance phase (greatest of the two) and the second occurs beyond the 60 percent point of the stance phase (Nigg et. al., 2003). The approximate values of frontal plane motion at heel-strike, the knee is abducted around 4 degrees and adducts to near 1 degree of abduction up through the first 30 percent. Following this peak in the adduction movement the knee then begins to abduct once again to around three degrees and then maintains this position from 50 to 70 percent of the stance phase before abducting back to the starting position (of heel strike) for toe-off.

With the intervention of an orthotic device there is a significant difference in the knee frontal plane range of motion than without the device. The knee at heel-strike is abducted closer to three degrees, then undergoes a reduced range of abduction motion up through the first 30 percent of the stance phase, peaking around zero degrees of abduction and adduction (relative neutral frontal plane alignment). The frontal plane motion of the knee with the use of an orthotic device follows a similar path as in the case without the device, except the abduction ranges are lesser in both the first and second peaks. In addition to the second abduction peak being lesser, it occurs later in the stance phase (McClay and Manal, 1998; Stackhouse et al., 2004).

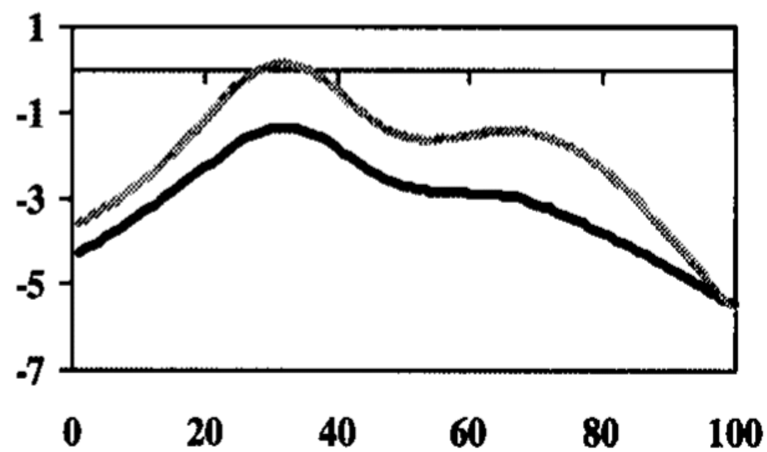

Figure 2.14 Graphical representation of knee joint kinematics the frontal plane as the knee begins in abduction and undergoes adduction. Range of motion is presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et al., 2004).

In the transverse plane, at heel-strike, the knee is externally rotated. As the stance phases progresses, the knee internally rotates beyond its neutral position orthotic device. The initial external range of motion value at the beginning of stance is around 12 degrees of external rotation. Rapidly the knee internally rotates through 30 percent of the stance phase, where internal rotation continues, at a reduced rate. Maximum knee internal rotation peaks around three degrees, occurring at close to 80 per cent of the overall duration of the stance phase. With the implementation of an orthotic device, the internal rotation motions about the knee joint during running were somewhat reduced. At the beginning of the stance phase was in a greater range of external rotation, around 15 degrees, and followed a similar pattern as the case without the use of an orthotic device. However there was reduction between two and three degrees in the amount of internal rotation between 30 and 80 percent of the duration of stance (Stackhouse et al., 2004).

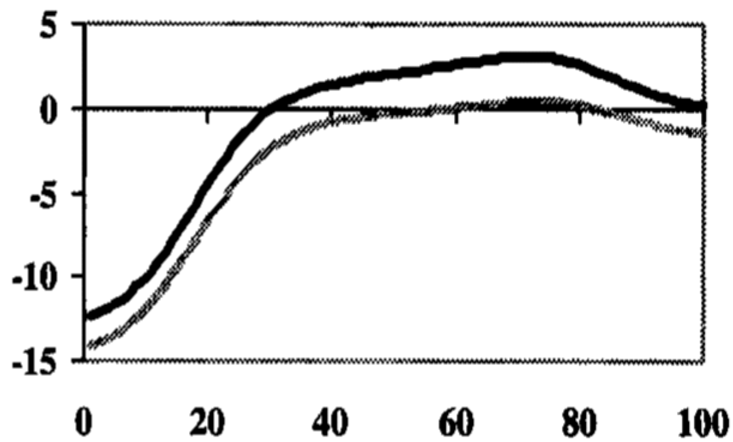

Figure 2.15 Graphical representation of knee joint kinematics in the transverse plane as the knee begins in external rotation and then internally rotates throughout stance. Range of motion is presented in degrees (vertical axis) as a function of the percent of stance phase (horizontal axis) in which these joint rotations occur. The darker line represents kinematic data from the normal running condition and the lighter line represents kinematic data when using an orthotic device (from Stackhouse et al., 2004).

2.10 Gender specific differences

As a clinician, anecdotally, it is notable that female runners are more frequently treated for overuse injuries about the lower extremity. Taunton et al. (2002) report female runners as being twice as likely to sustain a chronic overuse injury (stress fracture, iliotibial band friction syndrome, patellofemoral pain syndrome) from running compared to a male. One of the more notable and clinically measured differences between genders is the Q-angle, a representation of the line of pull on the quadriceps muscle on the patella via the line that intersects the anterior superior iliac spine and the patella . Females typically have a larger Q-angle. Heiderscheit, Hamill, and Caldwell (2000) looked at the influence of Q angle, In this study, no significant differences were noted between men and women in either of the measured kinematic parameters (maximum rearfoot eversion angle and maximum tibial internal rotation). It was reported, however, that the high Q angle group required more time for the tibia to reach maximum internal rotation compared to the low Q angle group.

2.11 Gender specific running kinematics

One intent of this study was to compare the effects of lower extremity kinematics between groups (with and without the use of an orthotic). Another was to contrast the kinematics between groups and gender. Since this study used a continuous method of collecting and analyzing data instead of using discrete kinematics (maximum joint excursions, time to peak motion, and velocities), a look comparison of the three dimensional kinematics of male and females runners is necessary. Ferber et al. (2003) performed such a study evaluating the mechanical differences in the lower extremity, hip and knee, between genders during running. Kinematic data reported in this study for the knee joint about the sagittal, frontal, and transverse planes were similar by comparison to the kinematics reported by Stackhouse and colleagues in 2004.

The kinematic data of Ferber et al. (2003) demonstrated that there is no reported difference between sagittal plane knee kinematics as a result of gender. Although it is noted that women do experience slighter higher ranges of knee flexion on average. This is in contrast to the finding that knee flexion angle for female subjects had been reported as 8º lower than males (Malinzak et. al., 2001).

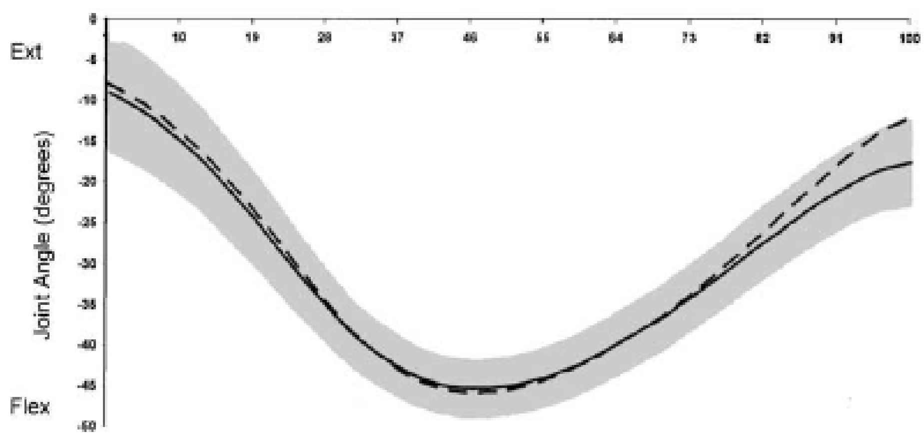

Figure 2.16 Comparison of sagittal plane running kinematics about the knee in male (solid line) subjects and female (dashed line) subjects. Joint angle (flexion – extension) is listed on the vertical axis while stance phases is normalized to a percentage of time on the horizontal axis. Shaded area represents $\pm$ 1 SD(From Ferber et al., 2003).

In the frontal plane, the pattern of motion between genders was similar. It is also noted that women experience greater frontal plane adduction of the knee (valgus) then men. There is not discussion in the report as to whether or not this difference was significant. This reports notes the difference to the findings of Malinzak et al.(2001) where females are shown to have on average 11º greater valgus direction or knee adduction.

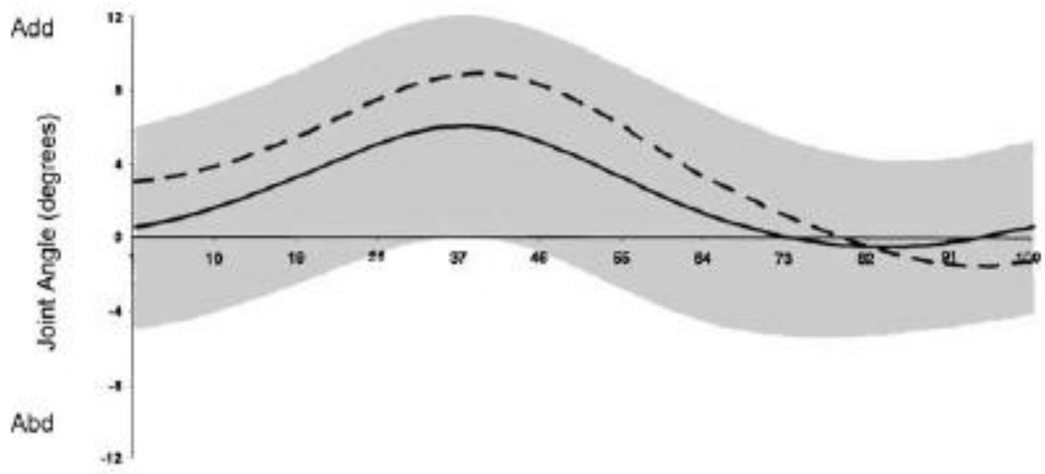

Figure 2.17 Comparison of frontal plane running kinematics about the knee in male (solid line) subjects and female (dashed line) subjects. Joint angle (adduction– abduction) is listed on the vertical axis while stance phases is normalized to a percentage of time on the horizontal axis. Shaded area represents ± 1 SD (From Ferber et al., 2003).

In the transverse plane female runners exhibited greater hip internal rotation at heel strike resulting in greater external rotation excursion compared to male runners. Again, the overall pattern of motion was similar between females and males.

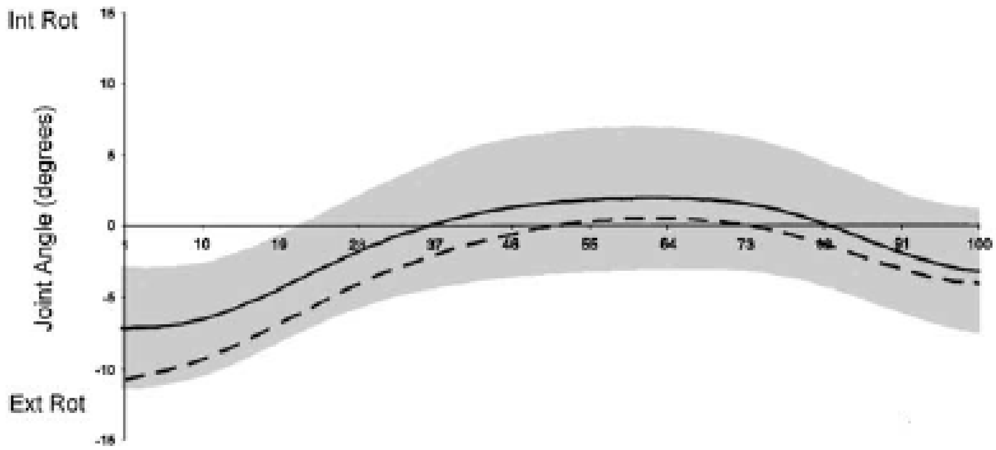

Figure 2.18 Comparison of transverse plane running kinematics about the knee in male (solid line) subjects and female (dashed line) subjects. Joint angle (adduction– abduction) is listed on the vertical axis while stance phases is normalized to a percentage of time on the horizontal axis. Shaded area represents ± 1 SD (From Ferber et al., 2003).

There is no kinematic data for the ankle in the case by Ferber et al. (2003) for kinematic comparison. There is an indication for the presence of movement variability occurring in the continuous measurement of running gait as noted by the standard deviation curves about the mean.

2.12 Variability

Human movement is inherently variable. The dynamic patterns of human locomotion allow for variance in cyclical movement patterns. This biologic variability may allow for greater understanding of the movement patterns in normal and pathologic populations. It is not yet fully understood at what level variability is either detrimental or beneficial. Based on a dynamical systems approach to human variability, the following have been suggested:

1) variability determines the stability of movement patterns around an attractor, large amounts of variability suggests unstable movement patterns, while small amounts of variability indicate stable movement patterns,

2) variability allows flexibility within the neuromuscular control system allowing for learning and adaptation to perturbations and,

3) provides beneficial perturbations so that appropriate movement patterns can be selected (James, 2004).

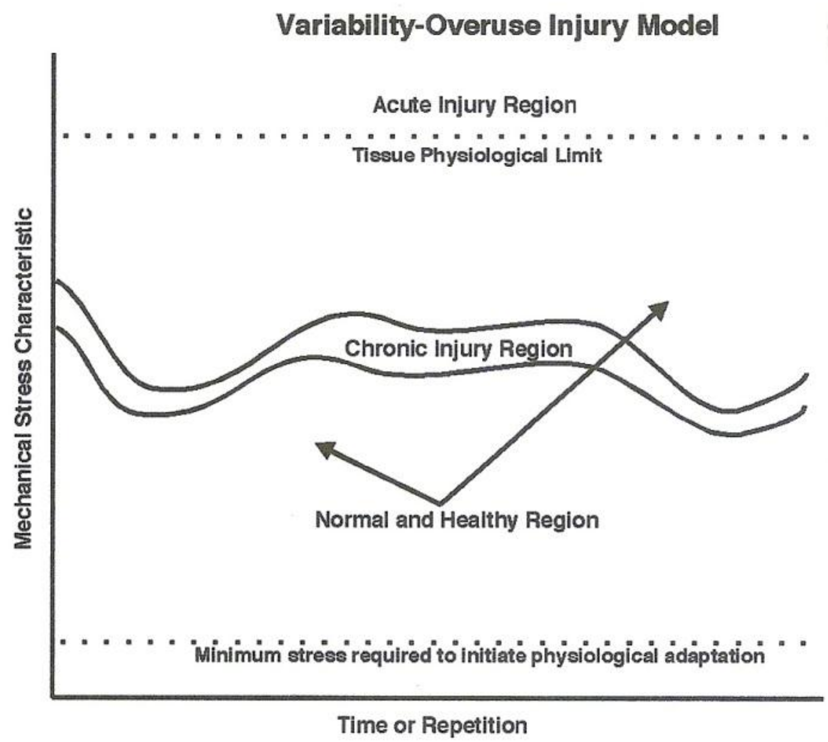

Figure 2.19 Variability-Overuse model of injury (from Stergiou, 2004).

In figure 2.19 a theoretical model relating variability and overuse injuries is displayed. Overuse injuries are submaximal stressors that occur through repetitive motions/actions overtime. An optimal range of variability may provide, in essence, a sense of relief from repetitive stress. Stergiou (2004) states that musculoskeletal health is maintained in a repetitive submaximal loading environment by variation of some critical value of the characteristics of loading (e.g. stress magnitude, frequency, direction). From a strict biomechanics perspective, movement variability makes sense for biological health to reduce and/or alter the loading pattern (compressive, tensional, frictional, torsional, etc.) acting upon the soft tissues – ligament, tendon, muscle fascia - of the lower extremity throughout repetitive motions. Too little variability and the stress would likely accumulate leading to overuse. From a central nervous system control perspective, too much variability could also be indicative of unstable movement. Recently, researchers have developed methods to learn more about movement variability, it's cause, and it's functional role, so it is now possible to begin understanding optimal variability among locomotive patterns (Stergiou et al, 2009).

Previous investigations into the effect of orthotic devices on running mechanics have looked at discrete measurements – max joint excursion(Bates et al., 1979; Munderman et al., 2003), peak joint velocity (Eng and Pierrynowski, 1994; Nawoczenski et al., 1995), joint angle at heel strike and toe off(Nigg et al., 2003; Novacheck, 1998). These past attempts at generalizing gait data used traditional, linear statistical methods – means and standard deviations - to quantify the results. Most recently, published studies (Ferber et al., 2003; and Nigg et al., 1997) have reported the amount of inter-subject and intra-subject variability of the result of orthotic interventions using continuous methods. Where kinematic data is expressed as an uninterupted series of data points (range of motion) over time (stance phase).

In a more recent study conducted by Kurz and colleagues (2002), a method for defining variability in locomotive patterns (the spanning set) was evaluated to determine whether or not this method was better at identifying variability compared to using standard deviations alone.

Larger standard deviation curves about the mean ensemble curve indicate greater variability; likewise, greater variability within the locomotive pattern will be indicated by a larger span between the vectors of a spanning set (Kurz et. al., 2002). The conditions compared in this study were barefoot versus shod running conditions, and the methods used to determine variability were coefficient of variation (CV) and mean deviation (MD) (both using standard deviations of the mean ensemble curve) and spanning sets. Even though the CV and MD did suggests increased amounts of variability, the differences were not statistically significant (Kurz et. al., 2002).

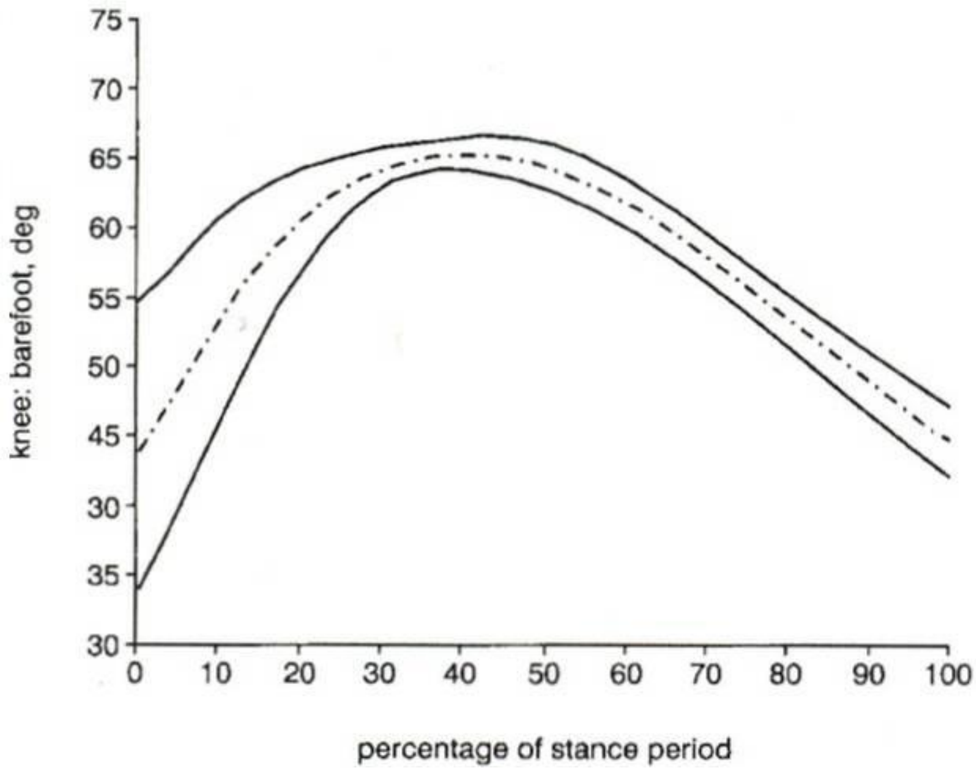

Figure 2.19 Sample mean ensemble curves (dotted lines) and standard deviations (solid lines). Ranges of motion of the knee – sagittal plane - in the barefoot condition are plotted along the vertical axis. Time in stance (from heel strike to toe off) is normalized to a percentage of time and plotted along the horizontal axis.(From Kurz et. al., 2002).

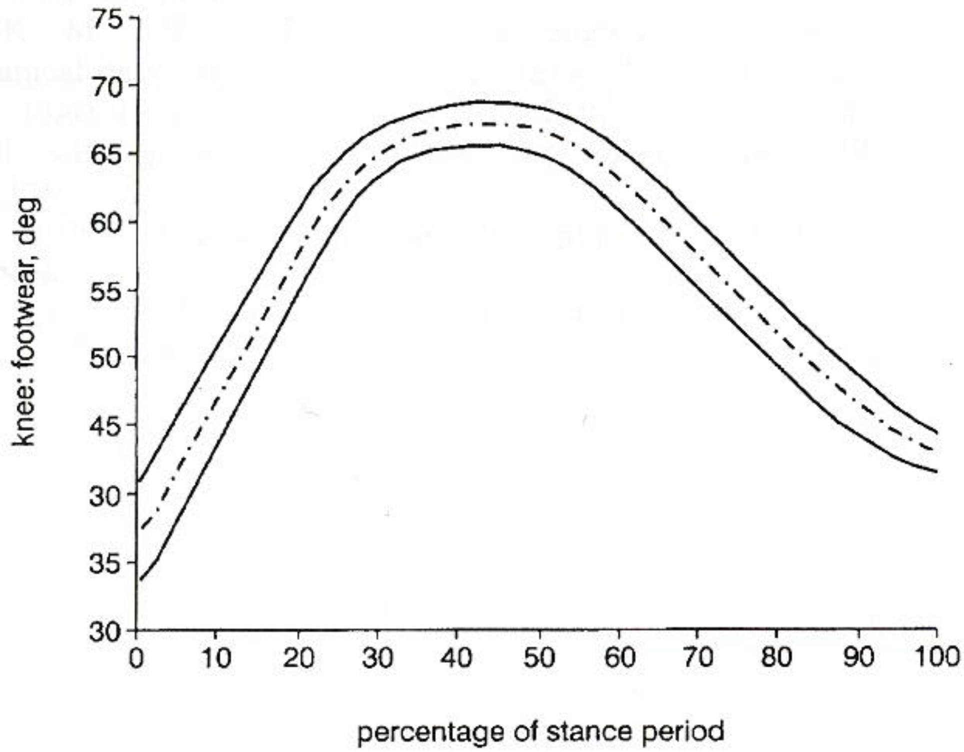

Figure 2.20 Sample mean ensemble curves (dotted lines) and standard deviations (solid lines). Ranges of motion of the knee – sagittal plane - in the footwear condition are plotted along the vertical axis. Time in stance (from heel strike to toe off) is normalized to a percentage of time and plotted along the horizontal axis.(From Kurz et. al., 2002).

In a similar study by Kurz and Stergiou (2003), the spanning set method was used to assess sagittal plane kinematic variability in the lower extremity (both ankle and knee joints) between a three conditions; running barefoot, running wearing a hard soled shoe, and running wearing a hard soled shoe. Significant differences for both joints were reported in this study between the barefoot and the hard shoe condition and the barefoot and soft shoe condition. No statistical differences were found between the variability between the two shoe conditions. This finding showing a reduction in variability as a result of wearing a shoe versus running barefoot might lead one to suspect that shoes may be more likely a causative factor in overuse running injuries.

Kurz and Stergiou (2003) also reported on the efficacy of the spanning set methodology to detect variations in sagittal plane kinematics during the stance phase of running while comparing changes in footwear type during treadmill running. Subjects were selected to run in a hard shoe, soft shoe, and barefoot. It was noted by the authors that the change in the mean ensemble curves for each running condition was an indication of the change in variability at each joint during stance, a finding that was similar to a followup study in 2004 which examined the influence of footwear on ankle coordination strategies which reported no significant difference between hard and soft shoes but a significant difference between both shod running to barefoot running. The barefoot running condition was shown to have a significantly larger amount of variability compared to the two other conditions. One possible theory of this effect is that barefoot running is more variable than shod running due to the diminished proprioceptive feedback provided by various types of footwear.

Even though spanning sets may be more sensitive to biological variability during locomotion - related to sensory information the foot receives during the stance period (Kurz and Stergiou, 2003; Kurz and Stergiou, 2004; Kurz et al, 2003) - it is not yet clear how variability can contribute to etiology of running related injuries. Basic kinematic variability does little to explain how the foot and ankle are coupled along with the lower leg and knee during the stance phase of running. The coupling patterns of the foot and ankle along with the knee during running have been investigated (Bates et. al., 1979; McClay and Manal, 1997) with little statistical significance found in the timing of coupling patterns between foot and ankle, lower leg, and knee motions. However, it has been reported that the timing of maximal foot, lower leg, and knee motions are more closely coupled in normal populations than in non-normal groups (McClay and Manal, 1997).

The spanning set offers one method of analyzing kinematic variability. There are other methods in use that follow a dynamic systems (continuous) approach. An example is the use of cumulative relative phase (CRP) portraits to look at variable changes in movement interactions between two segments during motion. Hamill et al. (199) used CRP measures to assess the variability of gait in healthy and pathologic patients suffering from patellofemoral pain (PFP) syndrome. Those with PFP were noted as demonstrating less movement variability.

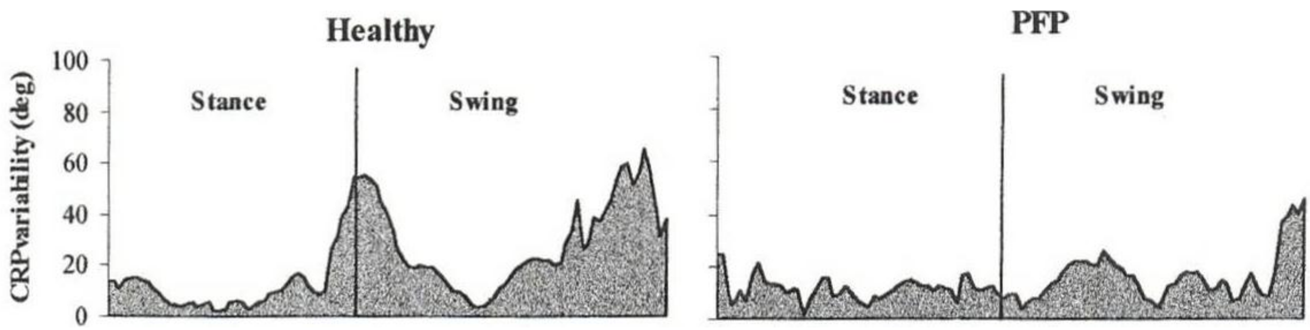

Figure 2.21 CRP data showing variability in movement across the stance and swing phase of gait in healthy and unhealthy (PFP) subjects. The shaded area under the curved line is presented as an indicator of variability in movement.

It does appear that dynamical systems analyses of motions, like running, are adequate to help create an understanding of the movement patterns. However, understanding the connection between varied movement patterns and underlying pathology and/or risk of mechanical failure requires greater analytical sensitivity. Biological variability is present in dynamic movement strategies. How this variability, in relation to improper mechanics and etiology of running injuries, can be quantitatively assessed is still not quite clear. More research is needed to help solidify current theories and findings.

Chapter 3

3.1 Data Collection

Thirty (15 male and 15 female) subjects age 18 to 40 years who were classified as recreational runners with pes planus (flat feet) participated in this study. Subjects were recruited from intercollegiate athletic programs, intramural athletic programs, and physical activity classes at the University of Kansas. Prior to participation in this study, all subjects were instructed to read and sign the required informed consent approved by the University of Kansas-Lawrence Human Subjects Committee. After providing informed consent, each subject underwent a pre-participation screening that included a health history questionnaire; individuals with cardiorespiratory/cardiovascular problems and/or musculoskeletal injuries within the past six months were excluded from participation in the study. The pre-participation screening included skeletal alignment measurements of leg lengths, Q-angle, and medial longitudinal arch heights.

Foot arch height measurements were assessed using a navicular drop test, which assesses hyper-pronation of the foot by measuring the displacement (change in height) of the navicular tuberosity as the foot goes from a non-weight bearing to a weight bearing position. Excess pronation is assumed to occur when the displacement of the navicular tuberosity is greater than 10 mm (Mueller et. al., 1993). Three consecutive measurements were taken and then averaged to calculate the arch height for each subject. In order to be classified as having pes planus and be eligible to participate in this study, subjects had to have an average navicular drop of 10 mm or greater. The pre-participation screening for all subjects was conducted by the same tester (a certified athletic trainer) to limit the occurrence of inter-tester error. The subjects were then scheduled to perform a treadmill running session in which they would run both with (orthotic) and without an orthotic device (control). Subjects were instructed to not exercise 24 hours prior to both data collection sessions.

Three-dimensional kinematic data were collected using a Visualeyez VZ3000 (Phoenix Technologies Inc., British Columbia, Canada) high speed motion measurement and tracking system. A wired marker system, consisting of light emitting diodes (LEDs) synchronized with the tri-camera system. Motion capture data were collected from each subjects' right lower extremity. Local joint coordinate systems were constructed on the lateral aspect of the right leg using three wired LEDs about the foot, ankle, shank, and thigh to calculate the relative displacements of each joint in each of the three cardinal planes. In addition, markers were placed on the posterior aspects of both the calcaneus and the thigh in a method similar to that used by McClay and Manal (1998) in an attempt to gain accurate kinematic data for rear foot motion.

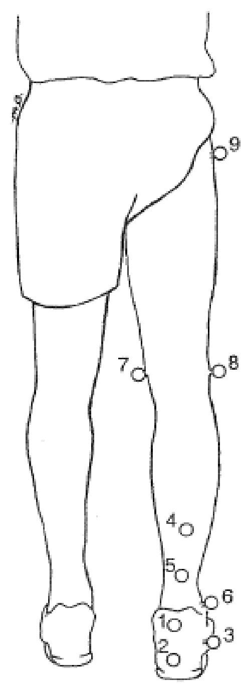

Figure 3.1 Demonstrating lower extremity marker placement (From McClay and Manal, 1998).

Figure 3.2 Image showing LED marker placements used to track motion of the lower extremity.

Figure 3.3 Image showing LED marker placement to track motion of the rearfoot.

Markers for the thigh and shank were attached directly to the subject’s skin using double sided tape (3M Inc.) and then secured using a self-adhesive athletic wrap (PowerFlex). Markers for the foot and ankle were secured directly to the subject’s shoes using underlying bony landmarks as reference points for marker locations. All subjects used the same style of “neutral” running shoe, the Nike Air Max Moto II (Beaverton, OR, USA).

Figure 3.4 Air Max Motto II shoe used by each subject.

The orthotic device used in this study was an over-the-counter, commercially marketed shoe insert, formerly marketed under the Flat Foot (Marathon Shoe Company, Boston, Massachussets, USA) brand name and now distributed as the Wedge Insole by Road Runner Sports (San Diego, CA, USA). This device is categorized as a soft orthotic and is made of PORON® cellular urethane. This product is designed to provide compensation for individuals having flat feet who are classified as hyper-pronators,. Specifically, these orthotic inserts are intended to keep feet in proper alignment during the mid-stance phase of gait by providing additional support along the medial longitudinal arch, resulting in proper skeletal alignment throughout the kinetic chain from the foot and ankle to the knee, hip, and back. The resultant effect of these orthotic devices, along with others, is to provide relief for many of the symptoms associated with overuse injuries related to physical activity.

Figure 3.5 Dorsal and plantar view of insert used for this study.

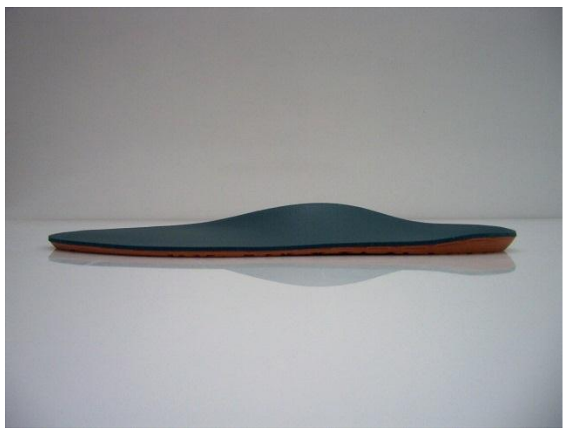

Figure 3.6 Lateral view of insert used for this study.

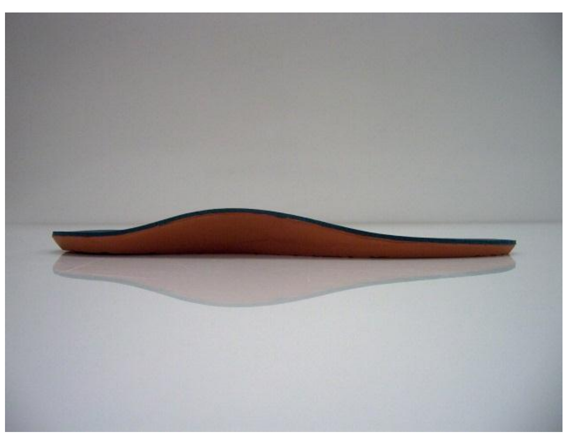

Figure 3.7 Medial view of insert used for this study.

All running trials were performed on a PreCor treadmill (Woodenville, Washington, USA). A global coordinate reference system was established along the base of the treadmill platform using three wired markers in view of the motion capture system prior to each subject's running trials. Each subject was fitted for the correct size of running shoe and then instructed to perform two, 15 min treadmill running protocols. The running protocol consisted of a five minute warm-up at a speed of 1.56 meters per second, then an initial running speed of 2.9 m/s, increasing over the next five minutes equally at one minute intervals until the final running speed of 3.35 m/s was reached. This final speed was maintained for the final five minutes of the treadmill protocol. Kinematic data were collected at 100 Hz for 60 s at the 12 min mark of the running protocol from the sagittal plane view and then again at 100 Hz for 60 s at the 14 minute mark of the running protocol from the frontal plane (posterior) view. The order of running trials (orthotic or no orthotic) were randomized between subjects. Each subject was given a 15 min rest period between running trials. The running protocol was then repeated following insertion or removal of the orthotic device, depending on the randomized order.

3.2 Data Reduction

Once the kinematic data was collected, it was then analyzed using VZAnalyzer Real-Time Motion Analysis Toolbox (Phoenix Technologies Inc., British Columbia, Canada). Using this software, rigid bodies were created for the thigh, shank, and ankle using the marker triads placed on the thigh, shank, and ankle for each subject for both the normal shod and orthotic conditions. This allowed for calculation of relative displacements between the joint segments resulting in the range of motion in all three planes (sagittal, transverse, and frontal) of motion for both the ankle and knee joints. In an attempt to quantify the actual amount of rear-foot motion between the calcaneus and the tibia, angular planes were created using the markers from the frontal plane views. All trials were filtered using a 4^{th} order low-pass Butterworth filter with a cutoff frequency of 6 Hz. All rigid body displacement and angular data were then imported into Excel (Microsoft Inc., Seattle, WA, USA) spreadsheet format for further analysis.

The filtered angular displacement data were then partitioned into stance phases for 50 consecutive foot strikes; it was assumed that heel strike during occurred in conjunction with maximal knee extension and that toe off occurred after a brief phase of flexion followed by extension. After the angular displacement data were partitioned into stance phases in all three planes of motion, relative joint angles were calculated using the rigid bodies of the ankle, shank, and thigh using the method described by Ferber et al. (2003). Relative angles of the knee and ankle were calculated using the angular displacement data of the thigh and shank (for the knee) and shank and foot (for the ankle). Examples of knee joint angular kinematic calculations in each plane of motion are as follows;

1. ϴ knee sagittal = ϴ thigh sagittal - ϴ shank sagittal, for knee flexion and extension,
2. ϴ knee frontal = ϴ thigh frontal - ϴ shank frontal, for knee abduction and adduction,
3. ϴ knee transverse = ϴ thigh transverse - ϴ shank transverse, for knee internal and external rotation.

Ankle joint kinematics calculated in the sagittal and transverse planes using the same equations as for the knee, except the angular displacement of the shank and foot were used, respectively. Frontal plane angular displacement data of the ankle joint, which was used to evaluate rear foot motion, was calculated using plane lines established along the posterior aspect of the tibia and the calcaneus.

Upon completion of angular kinematics for each joint segment of interest – ankle and knee, in all three planes of motion – sagittal, frontal, and transverse, for both conditions – control and orthotic, for both male and female, peak angular velocities and peak angular accelerations were observed along with the time to each peak angular value beginning at ground contact. It was reported in the literature that stance phase kinematic data of importance during running occurs within approximately the first 60 percent of the stance phase, or through mid-stance (Ferber et al, 2003; McClay and Manal, 1998).

3.3 Data Analysis

3.3.1 Traditional (Discrete) methods

To examine the amount of variability present in the kinematic data standard deviations were calculated for each period of the stance phase along with the coefficient of variation. Standard deviations (SD) of the kinematic data were calculated using the averages of each kinematic variable across the time domain of each data sample (50 stance phases, gait cycles) using Excel. Coefficients of variation (CV), which Stergiou (2004) expressed to be the most common quantity that represents a relative (normalized) variability measure, were calculated applying the following equation to each standard deviation data point.

$$CV = (SD/M) \times 100$$

3.3.2 Non-traditional (Dynamical Systems) method

Spanning set calculations were applied to the ankle and knee joint range of motion standard deviation curves using the following calculations:

1. $p(t) = \sum_{n=0}^{\infty} a_n t^n = a_0+a_1t+a_2t^2+\ldots$, for the above the mean curve and;
2. $g(t) = \sum_{n=0}^{\infty} b_n t^n = b_0+b_1t+b_2t^2 \ldots$, for the below the mean curve.

These above and below the mean curves represents time-series data that is ± 1 standard deviation about the mean.

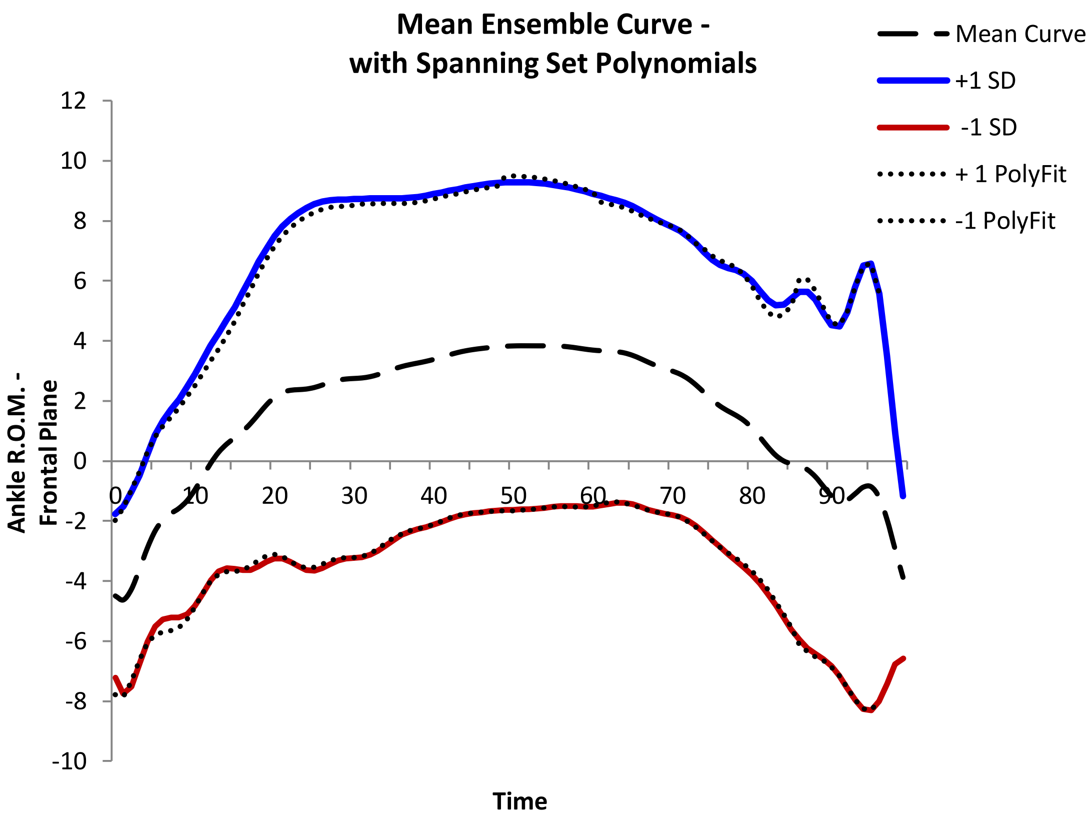

Figure 3.8 Graphic showing the continuous data set used for the spanning set analysis. Joint range of motion is listed along the vertical axis in degrees. The horizontal axis represents stance phase normalized to a percentage of time. Mean data presented by the dashed black line. SD curves in solid red and blue lines represent the deviations 1 SD above and below the mean, respectively. Each black line is the polynomial line of best fit for each SD curve.

Once the polynomials were calculated using the above methods, the spanning set was determined from each formula using the model.

$$f = \begin{matrix} \mathrm{a}0 & b0 \\ . & . \\ an & bn \end{matrix}$$

And then calculated using the equation below where "u" represents the coefficients of the first vector above and "v" represents the coefficients of the second vector above.

$$y = \|u - v\|$$

In this case, the larger the spanning set, the greater the variability in the joint range of motion pattern.

<u>Spanning set data example;</u>

Poly Fit + = 6.493343 – 4.07048t + .66324t2 -.04043t3 +.001212t4 -1.9E-05t5 +1.5E-07t6 -4.7E-10t7

Poly Fit - = -9.760 + 1.17091t – 1.08E-1t2 + 5.49E-03t3 -1.52E-04t4 +2.29E-06t5 -1.77E-08t6 + 5.47E-11t7

Using the coefficients from the polynomials, the vector math was set up using the matrix of:

PF+ - PF- =

$$\begin{vmatrix} 6.493343 \\ -4.07048 \\ 0.66324 \\ -0.04043 \\ 0.00121 \\ -1.90E-05 \\ 1.50E-07 \\ -4.70E-10 \end{vmatrix} - \begin{vmatrix} -9.76 \\ 1.17091 \\ -1.08E-01 \\ 5.49E-03 \\ -1.52E-04 \\ 2.29E-06 \\ -1.77E-08 \\ 5.47E-11 \end{vmatrix} = \begin{vmatrix} 16.253343 \\ -5.24139 \\ 0.77124 \\ -0.04592 \\ 0.001362 \\ -0.00002129 \\ 1.677E-07 \\ -5.247E-10 \end{vmatrix}$$

A spanning set was then calculated using the difference between the two vectors (far right above) using the following calculation:

$$\|u - v\| = \sqrt{(16.253343)^2 + (-5.24139)^2 + (.77124)^2 + (-.04592)^2 + (.001362)^2 + (-.00002129)^2 + (1.677E-07)^2 + (-5.247E-10)^2}$$

$$= 17.09504$$

The value of the spanning set can then be compared to the SDs and CVs to determine the relative magnitudes of the kinematic variability found at the ankle and knee joints for each subject. Using SPSS (SPSS, Inc.; Chicago, IL, USA), an analysis of variance was used to assess the effects of calculation method (SD, CV, or spanning set) on orthotic- and gender-related kinematic variability of the ankle and knee joint range of motion kinematic data.

Chapter 4

Data analysis began looking at the raw kinematic data of both the ankle and knee specifically in the three planes of motions (transverse, frontal, and sagittal) for both the male and female subjects running with or without the selected orthotic. Following is the presentation of this data as it pertains to the ranges of motion measured from the point at which the foot came into contact with the treadmill (foot strike) until a change in peak motion was seen (mid-stance) indicating the onset of propulsion. The ankle data is presented first followed by the knee keeping the planes in the order listed above. This chapter presents the range of motion data in the sagittal, transverse, and frontal planes of motion for both the ankle and knee joints as a function of gender and orthotic. In addition, three different methods (standard deviation, coefficient of variation, and spanning set) for calculating movement variability of the range of motion data are compared.

4.1 Kinematic Data

4.1.1 Ankle joint range of motion

Range of motion values in the transverse plane were $12.3 \pm 5.13°$ and $12.9 \pm 4.1°$ with the orthotic and $13.4 \pm 4.6°$ and $13.3 \pm 5.8°$ and in controls, in males and females, respectively (see Figure 4.1). In the transverse plane, there were no significant differences due to the main effects of orthotic condition, gender, as well as the interaction effect between orthotic condition and gender.

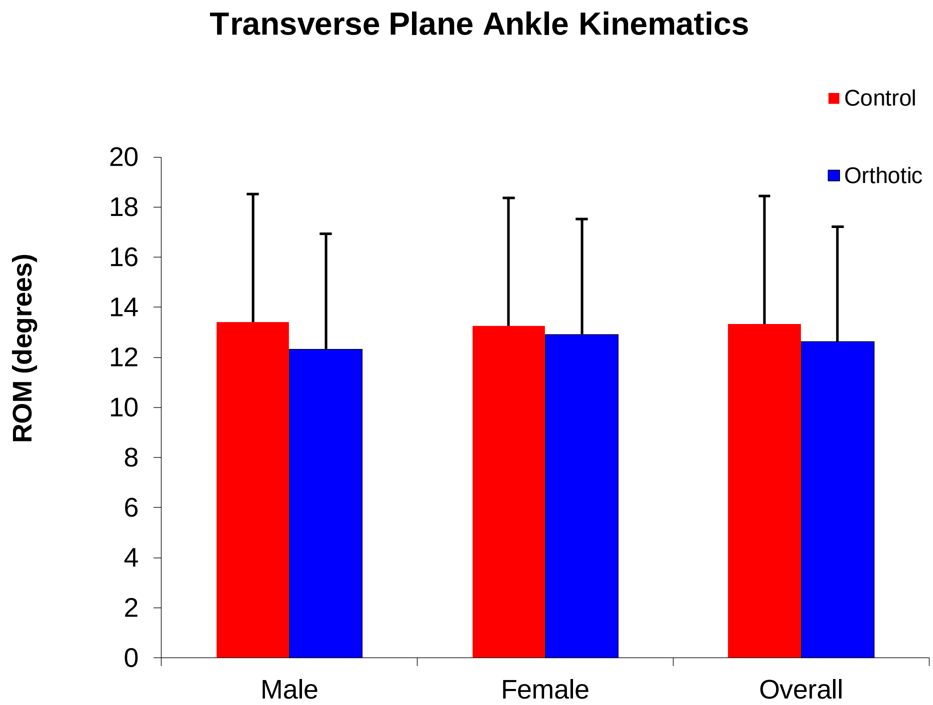

Figure 4.1 Mean kinematic data of the ankle joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation.

Range of motion values in the frontal plane were 14.6 ± 3.4° and 14.0 ± 4.1° with the orthotic and 13.8 ± 5.2° and 13.8 ± 4.0º in controls, in males and females, respectively (see Figure 4.2). In the frontal plane, there were no significant differences due to the main effects of orthotic condition and gender, gender, as well as the interaction effect between orthotic condition and gender.

Frontal Plane Ankle Kinematics

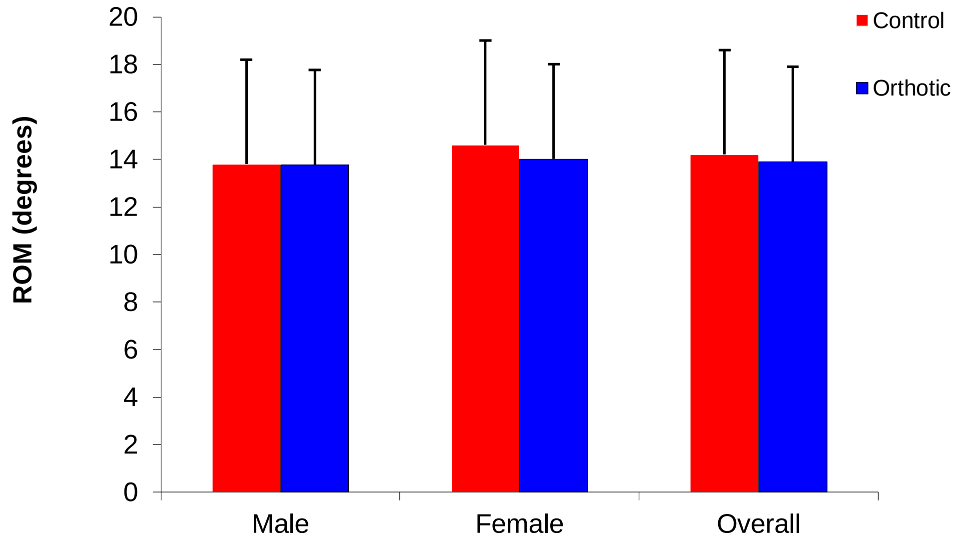

Figure 4.2 Mean kinematic data of the ankle joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation.

Range of motion values in the sagittal plane were 17.7 ± 8.9° and 18.2 ± 6.6° with the orthotic and 17.3 ± 5.4° and 18.2 ± 6.6º in controls, in males and females, respectively (see Figure 4.3). In the sagittal plane there were no significant differences due to the main effects of orthotic condition, gender, as well as the interaction effect between orthotic condition and gender

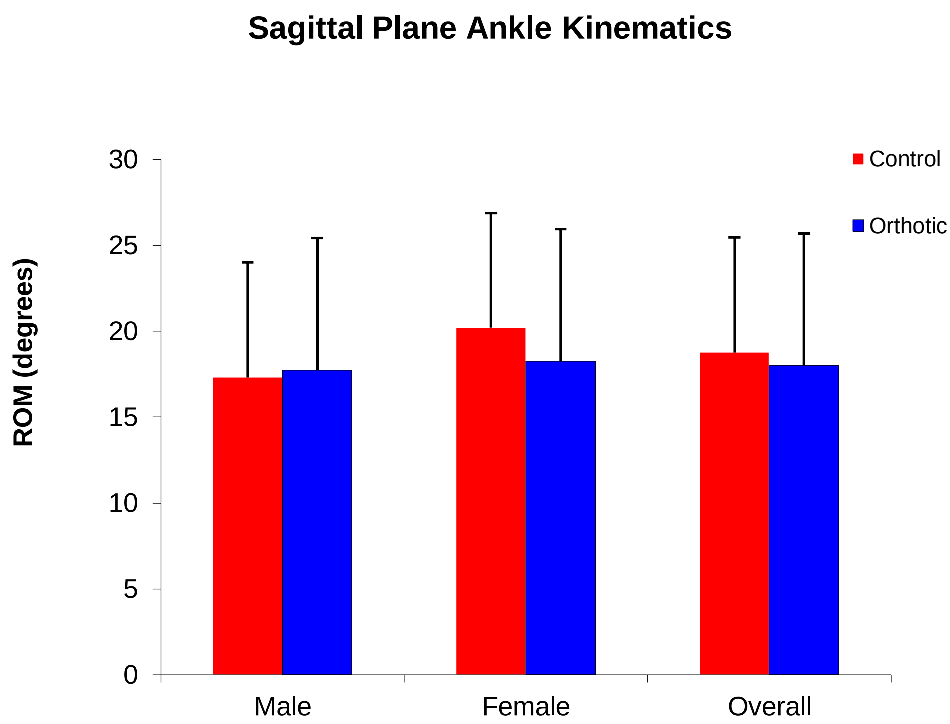

Figure 4.3 Mean kinematics data of the ankle joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation.

4.1.2 Knee joint range of motion

Range of motion values in the transverse plane were 10.7 ± 2.8° and 13.1 ± 3.9° with the orthotic and 11.3 ± 2.8° and 13.1 ± 3.9º in controls, in males and females, respectively(see Figure 4.4). In the transverse plane, there were no significant differences due to the main effects of orthotic condition, gender, as well as the interaction effect between orthotic condition and gender.

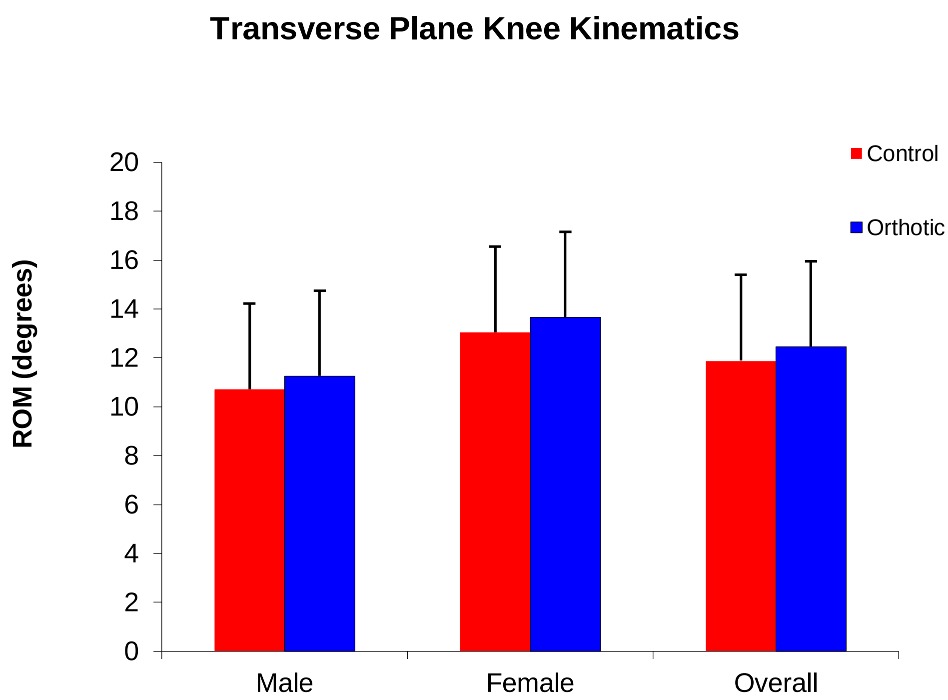

Figure 4.4 Mean kinematics data of the knee joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation.

Range of motion values in the frontal plane were 9.1 ± 1.9° and 10.6 ± 4.4° with the orthotic and 9.9 ± 1.9° and 9.7 ± 5.3º in controls, in males and females, respectively (see Figure 4.5). In the transverse plane, there is significance due to the interaction effect between orthotic condition and gender ($F = 4.544$, $P <.05$) and gender ($F = .263$, $P < .001$). There were no significant differences due to the main effects of orthotic condition.

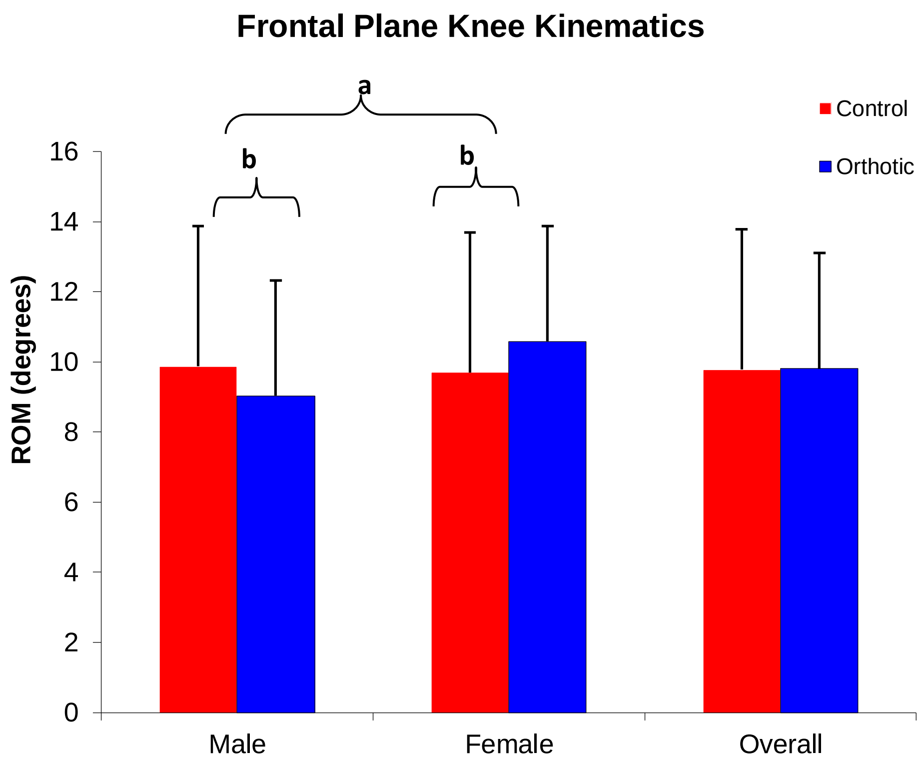

Figure 4.5 Mean kinematics data of the knee joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation. "a", Indicates significant difference between the interaction effect of orthotic and gender. "b", Indicates significant difference due to gender.

Range of motion values in the sagittal plane were 21.9 ± 4.1° and 20.9 ± 3.5° with the orthotic and 23.6 ± 4.1° and 23.1 ± 4.4º in controls, in males and females, respectively (see Figure 4.6). In the sagittal plane, there was a significant difference due to the main effects of orthotic condition ($F = 11.06$, $P < .05$). There were no significant differences due to the interaction effect between orthotic condition and gender, or gender.

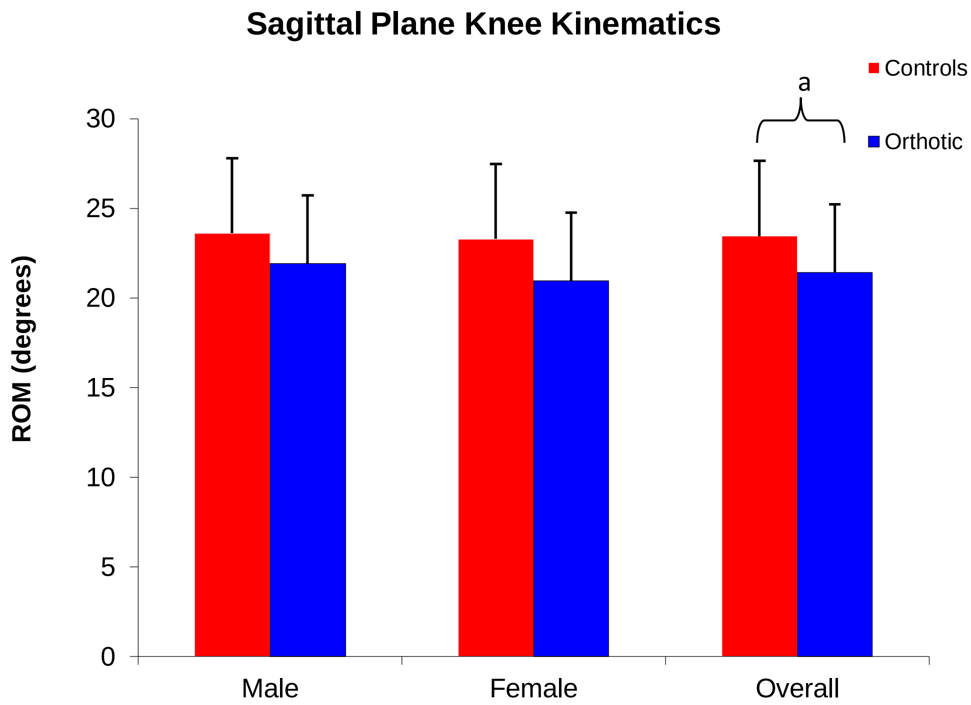

Figure 4.6 Mean kinematics data of the knee joint between subjects. Vertical axis represents maximum average joint range of motion. Horizontal axis compares the effects of footwear condition on gender and overall. Error bars represent standard deviation. “a”, Indicates significant difference due to the main effect of the orthotic.

Change scores of frontal plane (Figure 4.7) ankle motion (from inversion to eversion as the foot accepts load from foot strike to mid-stance) were created by comparing the range of motion with and without the use of the orthotics. The difference between the two conditions provided the change score. The results of this showed the variable response to the orthotic device among subjects. With the device, some subjects showed an increase in rearfoot motion and some showed a decrease in rearfoot motion. The degree of change was also variable.

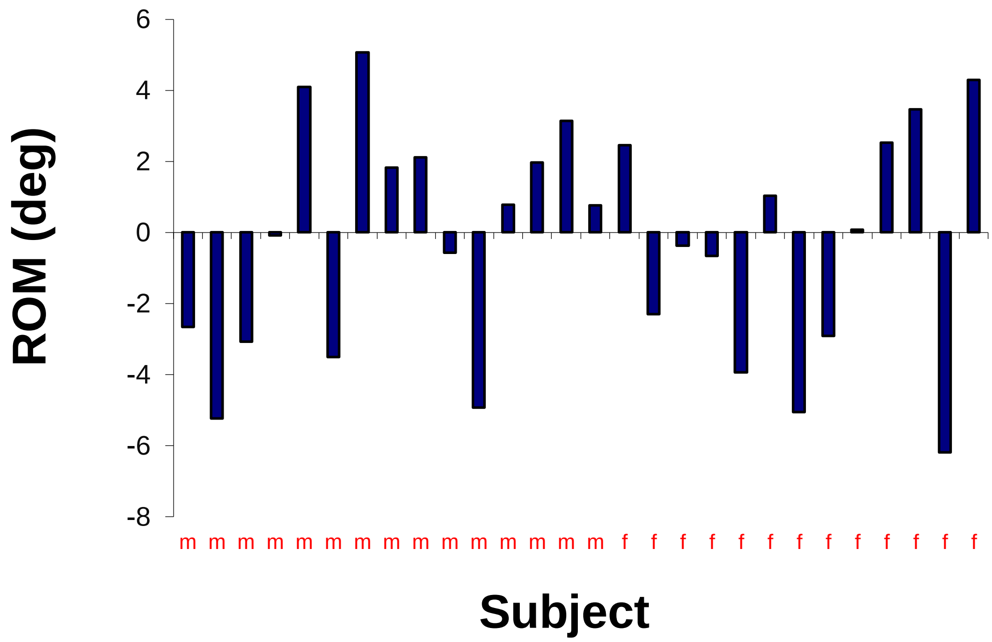

Figure 4.7 Indication of the between subject variable response upon rearfoot mechanics as a result of the use of the orthotic. The difference in range of motion of the forefoot with and without the use of the orthotic is plotted along vertical axis. The horizontal axis shows the subject by subject comparison as well as indicating gender (“m” = male, “f” = female).

4.2 Kinematic Variability

4.2.1 Kinematic variability of the ankle.

Variability in the transverse plane standard deviation (SD) revealed a significant difference due to the main effects of orthotic condition ($F = 22.644$, $P < .05$). There was not significance as an interaction effect between orthotic condition and gender or gender. Looking at the coefficient of variation (CV) in the transverse plane there was a significant difference due to the main effects of orthotic condition ($F = 6.976$, $P \leq .05$). There was not significance as an interaction effect between orthotic condition and gender or gender. For the spanning set (SS) in the transverse plane there was not a significant difference due to the main effects of orthotic condition ($F = 9.547$, $P < .05$). There was not significance as an interaction effect between orthotic condition and gender or gender.

Ankle
Transverse Plane

	No Orthotic			Orthotic		
	SD	CV	SS	SD	C.V	SS
Male (15)	13.42	14.08	19.14	12.34	27.23	30.85
(±)		(5.59)	(7.60)		(21.30)	(8.50)
Female (13)	13.26	15.76	20.29	12.92	23.13	24.97
(±)		(10.81)	(10.15)		(17.03)	(8.50)
Overall (28)	13.35 a	14.86 b	19.67 c	12.6 a	25.32 b	28.12 c
(±)		(8.30)	(8.72)		(19.80)	(13.79)

Figure 4.8 Charts the transverse plane ankle variability data. Significant differences due to the main effect of the orthotic noted using “a” for SD, “b” for CV, and “c” for SS.

Variability in the frontal plane SD revealed a significant difference due to the main effects of orthotic condition ($F = 5.392$, $P = <.05$) and gender ($F = 4.827$, $P <.05$). There was no significance as an interaction effect between orthotic condition and gender. Looking at the CV in the frontal plane there was not a significant difference due to the main effects of orthotic condition, as an interaction effect between orthotic condition and gender, or gender. For the SS in the frontal plane there was a significant difference due gender ($F = 4.492$, $P < .05$). There was no significant difference due to the main effects of orthotic condition or as an interaction effect between orthotic condition and gender.

Ankle
Frontal Plane

	No Orthotic			**Orthotic**		
	S.D.	C.V.	S.S.	S.D.	C.V.	S.S.
Male (15)	13.80 b	15.95	21.69 c	12.34 b	17.30	23.66 c
(±)		(9.26)	(12.60)		(7.43)	(10.17)
Female (13)	14.60 b	10.03	13.40 c	12.92 b	14.41	19.40 c
(±)		(5.25)	(7.01)		(7.00)	(9.43)
Overall (28)	14.17 a	13.20	17.84	12.61a	15.96	21.68
(±)		(8.11)	(11.04)		(7.25)	(9.89)

Figure 4.9 Charts the frontal plane ankle variability data. Significant differences due to the main effects of the orthotic indicated using “a”. Significant difference due to gender indicated using “b” for SD, and “c” for SS.

Variability in the sagittal plane SD revealed no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. Looking at the CV in the sagittal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. For the SS in the sagittal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender.

Ankle
Sagital Plane

	No Orthotic			**Orthotic**		
	S.D.	C.V.	S.S.	S.D.	C.V.	S.S.
Male (15)	17.32	13.18	17.92	17.74	19.51	26.68
(±)		(5.42)	(7.37)		(10.65)	(14.56)
Female (13)	20.20	16.28	21.76	18.24	20.27	27.28
(±)		(12.95)	(17.35)		(16.28)	(21.92)
Overall (28)	18.66	14.62	19.70	17.97	19.86	26.96
(±)		(9.62)	(12.87)		(13.30)	(17.99)

Figure 4.9 Charts the sagittal plane ankle variability data.

4.2.2 Kinematic variability of the knee.

Variability in the transverse plane SD revealed a significant difference due gender ($F = 4.439$, $P < .05$). There was no significance to the main effects of orthotic condition or as an interaction effect between orthotic condition and gender. Looking at the CV in the sagittal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. The SS in the transverse plane revealed a significant difference as an interaction effect between orthotic condition and gender ($F = 5.306$, $P < .05$). There was no significance to the main effects of orthotic condition or gender.

Knee
Transverse Plane

	No Orthotic S.D.	C.V.	S.S.	**Orthotic** S.D.	C.V.	S.S.
Male (15)	10.73 a	15.55	17.37 b	11.25 a	13.12	17.86 b
(±)		(9.16)	(10.23)		(6.99)	(7.22)
Female (13)	13.06 a	16.83	17.84 b	13.66 a	15.53	22.67 b
(±)		(13.16)	(7.58)		(7.30)	(10.65)
Overall (28)	11.81	16.15	17.59 b	12.37	14.23	20.09 b
(±)		(11.00)	(8.94)		(7.10)	(9.14)

4.10 Charts the transverse plane knee variability data. Significant difference due to gender indicated using “a”. Significant difference due to the interaction effect of the orthotic on gender indicated using “b”

Variability in the frontal plane SD revealed no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. Looking at the CV in the frontal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. For the SS in the frontal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender.

Knee
Frontal Plane

	No Orthotic			Orthotic		
	S.D.	C.V.	S.S.	S.D.	C.V.	S.S.
Male (15)	9.87	16.80	18.77	9.03	22.61	24.24
(±)		(7.88)	(8.80)		(23.31)	(11.12)
Female (13)	9.70	29.81	35.69	10.58	22.74	33.21
(±)		(24.26)	(29.46)		(14.91)	(21.77)
Overall (28)	9.79	22.84	26.62	9.75	22.67	28.40
(±)		18.58	(22.36)		(19.50)	(17.19)

Figure 4.11 Charts the sagittal plane ankle variability data.

Variability in the sagittal plane SD revealed no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. Looking at the CV in the sagittal plane there was no significant difference as an interaction effect between orthotic condition and gender, due to the main effects of orthotic condition, or due to gender. For the SS in the frontal plane there was a statistically significant difference due to the main effects of orthotic condition ($F = 17.36$, $P < .001$). There was no significant difference as an interaction effect between orthotic condition and gender or due to gender.

Knee
Sagittal Plane

	No Orthotic			Orthotic		
	S.D.	C.V.	S.S.	S.D.	C.V.	S.S.
Male (15)	23.60	8.02	8.96	21.91	9.81	14.32
(±)		(1.34)	(1.50)		(2.74)	4.00
Female (13)	23.28	9.28	11.10	20.94	9.62	14.05
(±)		(6.11)	(7.32)		(2.76)	(4.03)
Overall (28)	23.45	8.61	9.96 a	21.46	9.72	14.20 a
(±)		(4.24)	(5.11)		(2.70)	(3.94)

4.12 Charts the transverse plane knee variability data. Significant differences due to the main effects of the orthotic indicated using "a".

4.3 Summary of Results

Kinematic Data

- Knee Joint Range of Motion
  - Frontal Plane
    - Interaction effect of orthotic condition and gender (F = 4.544, P <.05)
    - Difference due to gender (F = .263, P < .001)
  - Sagittal Plane
    - Difference due to main effects of orthotic condition (F = 11.06. P < .05)

Kinematics Variability Data

- Ankle Joint Kinematic Variability
  - Transverse Plane; Significance due to the main effect of orthotic condition upon:
    - SD = (F = 22.644, P < .05)
    - CV = (F = 6.976, P < .05)

- SS = ($F = 9.547$, $P < .05$)

- Frontal plane, Significance due to:
    - SD - Main effects of condition ($F = 5.392$, $P < .05$)
    - SD -Gender ($F = 4.827$, $P < .05$)
    - SS - Gender ($F = 4.492$, $P < .05$)

- Knee Joint Kinematic Variability
    - Transverse Plane; Significance due to:
        - SD - Gender ($F = 4.439$, $P < .05$)
        - SS - Interaction effect between orthotic condition and gender ($F = 5.306$, $P < .05$)
    - Sagittal plane; Significance due to:
        - SS - Main effects of condition ($F = 17.36$, $P < .001$)

Chapter 5

The purpose of this study was to investigate lower extremity kinematics and kinematic variability in male and female recreational runners with pes planus (flat feet) when using an over-the-counter (OTC) soft orthotic device during treadmill running. In this chapter, the angular displacement (range of motion) results for both the ankle and knee joints in the sagittal, transverse, and frontal planes will be discussed and compared with previous research findings. In addition, the use of two traditional methods (standard deviation and coefficient of variation) and one non-traditional method (the spanning set) for calculating kinematic variability will be assessed.

5.1 Lower Extremity Kinematics

It is noted in a study by Eng and Pierrynowksi (1994), a similar trend of ankle motion in the frontal and transverse planes being reduced (as an effect of the use of an orthotic device) with an increase in motion about the knee throughout the first 50% of stance (contact to mid-stance) an effect similar to our kinematic data. The data analyzed for the kinematic portion of this study is consistent with previous studies (Eng and Pierrynowksi, 1994; Ferber et al., 2003; McClay and Manal, 1998; Nawoczenski et. al. 1995, Stackhouse et al., 2004) examining kinematic data about both the ankle and knee in multiple planes of motion in runners.

5.1.1 Effects of gender on lower extremity kinematics

About the ankle in the transverse, frontal and sagittal planes there were no significant differences due to gender alone. This finding is consistent to those of previous authors (Eng and Pierrynoski, 1994; and Nawoczenski et al., 1995) where no differences were found as a result of gender. Other studies (Heiderscheit et al., 2000; Pink, 1994) did report differences among genders, especially in the frontal planes, where females presented with higher peak joint excursions.

About the knee in the transverse, frontal and sagittal planes there were no significant differences due to gender alone. The kinematics presented in this study were consistent with the kinematics findings of Ferber et al. (2003) and McClay and Manal (1998) about the transverse and frontal planes. The only difference in our findings is Ferber et al. (2003) reported females having greater sagittal plane knee joint range of motion compared to males. Our data suggests the groups are no different.

5.1.2 Effects of orthotics on lower extremity kinematics

About the ankle in the transverse, frontal and sagittal planes there were no significant differences due to the effect of the orthotic. Even though the finding was not significant, it appeared that the orthotic device affected, foot pronation throughout the weight bearing phase of running up to propulsion. However, this effect was not consistent across subjects or groups of subjects. Some subjects were noted as “responders” while others were noted as “non-responders”. This finding is consistent with those of Nawoczenski et al. (1995).

About the knee in the transverse and frontal planes there were no significant differences due to the effect of the orthotic. In the sagittal plane, an overall difference (23.44º to 21.43) due the use of the orthotic presented at a level of significance. In the case where the ankle presented with greater stiffness as a result of running with the orthotic it is apparent that mobility may indeed be enhanced at the knee which might also indicate greater hip mobility as well. There did appear to be an adaptive response which involved greater knee mobility in the transverse and frontal plane, potentially as a result of greater rigidity (less mobility) present in ankle kinematics when using the orthotic. It is possible when coupled with less ankle mobility greater mobility is likely to occur about the knee in a given plane. It is not clear at this time whether any clear definition between physiologic response and pathologic response to the use of the orthotic device can be assumed. Nigg (2001) challenged the modern use of running shoes, orthotic devices, and inserts as well as skeletal alignment suggesting the effects seen among the variables "produce only small, not systematic, and subject-specific changes of foot and leg movement". Similar to these findings, these data suggest that the change in lower extremity kinematics were specific to each subject and not systematic. Therefore, it is hard to predict any benefit or detriment to the runner.

5.1.3 Interaction effects of gender and orthotics on lower extremity kinematics

The only significant interaction effect between gender and orthotic occurred in the frontal plane kinematics about the knee. Male range of motion decreased slightly (9.87º to 9.03º) due to the use of the orthotic. Female range of motion increased slightly (9.69º to 10.59º). This seems consistent with previous findings (Ferber et al, 2003; Malinzak et al., 2001) suggesting females are likely to experience more frontal plane range of motion about the knee.

5.2 Lower Extremity Kinematic Variability

Based on the work of Stergiou and Decker (2011), an "optimal" amount of variability in healthy movement is not clearly understood. It is assumed that since none of the subjects who participated in this study were injured that the data is representative of a healthy system responding to changes in sensory feedback as a result of footwear condition. In previous research by Stergiou et al. (2006), only two case studies were presented on the concept of optimal variability; they simply presented a theoretical framework for discussion and not a direct application to a large comparative analysis such as this involving a larger sample population with varying degrees of running ability.

There was an observable effect of both inter- and intra-subject variability. Looking at inversion change scores (figure 4.7) about the frontal plane of the ankle it was apparent that in both male and female groups there is an inconsistent response to the application of the orthotic device when running. A few subjects showed little to no change between conditions. Some subjects presented with greater motion in this plane while running with the orthotic while others responded with a reduction in frontal plane motion, a finding consistent with Nawoczenski et al. (1995). This variability is often overlooked when means are evaluated as an overall affect revealing minimal effect among groups or subjects overall. Upon further evaluation it was considered that discrete (means of point specific data) measures of variability (standard deviation, coefficient of variation) may not be as sensitive to variability occurring in healthy movement patterns when compared to dynamic measures of variability (the spanning set) occurring continuously over time. Kurz and Stergiou (2003) have explored these measures previously and have found the spanning set methodology to offer greater sensitivity to movement variability analysis.

5.2.1 Effects of gender on lower extremity kinematic variability

About the ankle in the transverse plane there were no significant differences among the three measures of variability as an effect of gender. A finding consistent with those of Ferber et al. (2003).

About the knee in the transverse plane there was a significant effect due to gender in both the SD and SS measures. Male SD (10.73 and 11.25) was lower when compared to female SD

(13.06 and 13.66). A similar effect was observed between genders for the SS measure in males (17.37 and 17.86) and females (17.84 and 22.67). There are no significant differences or notable trends in frontal or sagittal plane variability. Although Ferber et al. (2003) did not report any significant findings, it was noted that females had a tendency for greater joint kinematic variability compared to males.

5.2.2 Effects of orthotics on lower extremity kinematic variability

About the ankle all three measures of variability present with a level of statistical significance in the transverse plane due to the conditional effect of the orthotics. With orthotic SD (12.6) decreased compared to without (13.35). With orthotic CV (25.32) and SS (28.12) increased from CV (14.86) and SS (19.67) without. It is unclear as to whether or not this difference between variability measures is an indicator of enhanced sensitivity. In the frontal plane only SD revealed significance in groups with (12.61) and without (14.17) orthotics. As was previously noted, this decrease in variability was in contrast to the increase in CV (15.96 from 13.20) and SS (21.68 and 17.84) with the use of the orthotic. Again, it is not clear whether this change is an indicator of greater sensitivity to variability. In the sagittal plane there are no reportable differences or consistent trends in patterns of variability. As noted by Kurz and Stergiou (2002) there were no significant differences reported as a result of footwear conditions. Our data suggests, this case, there may be an effect on variability due to the orthotic.

About the knee there are no significant differences or trends as a result of the orthotic on SD, CV, or SS in the transverse or frontal plane. In the sagittal plane there is a significant effect of the orthotic as the SS increased to 14.20 with the orthotic from 9.96 without. The CV also indicated an increase in variability with (9.72) orthotic compared to without (8.61) orthotic.

However this was not significant. Each of these measures was in contrast to a decrease in variability indicated by the SD with (21.46) orthotic compared to without (23.45). Again, these findings suggest an enhancement in variability as a result of the orthotic, running counter to the idea proposed by Kurz and Stergiou (2002, 2004) that footwear reduces the presence of variability while running. Hamill et al. (1999) report less variability in unhealthy subjects which might indicate our data is presented within an "optimal" range of variability.

5.2.3 Interaction effects of gender and orthotics on lower extremity kinematic variability

About the ankle in the transverse, frontal, and sagittal planes there are no significant differences as an interaction effect of gender and orthotic upon SD, CV, or SS. In this case the SS did not present with greater sensitivity to movement variability.

About the knee in the transverse plane there is a significant interaction effect among SS data. Male SS with orthotic (17.86) was slightly higher than without (17.37. Female SS with was much higher with orthotic (22.67) compared to without (17.84). In the frontal and sagittal planes there are no interaction effects to report. This change might be indicative of the changes in ankle joint kinematics and variability as noted by (Ferber et al. 2003; McClay and Manal, 1998).

5.3 Conclusion

It is not clear based on this data if one could clearly suggest or refute the use of an OTC orthotic, such as the one used in this study, for direct control of ankle and knee mechanics. There clearly is a subjective individualized response. For some the less expensive OTC device may provide an inexpensive alternative to a higher priced custom device. For others there may be a definite need for a more custom fit appliance. An interesting note, Kurz and Stergiou

(2004) found there is little difference in the movement variability and coordinative strategy between a hard or a soft shoe whereas each of the shoe wear conditions is significantly different compared to either shod condition and running barefoot. It is a likely explanation that this may be the case in our study since we used a shoe designed for running combined with an over the counter soft orthotic. Also, the adaptive changes seen among the subjects in this study are presumed to be that of healthy subjects. There is no reason to suspect the changes in kinematics and variability reported in this study are the effect of a pathologic (abnormal) control strategy. Each subject may have been operating within a healthy range of variability.

It is not completely clear as to whether or not the SS method used in this study is clearly more sensitive to movement variability although when significant differences were noted SS was included. The variability data analyzed in this study was from filtered kinematics data whereas the method described by Kurz and Stergiou (2003) used unfiltered kinematic data.

5.4 Future Directions

Continuous measures of kinematic data and variability are likely better suited for the analysis of gait data. More research is needed comparing the three dimensional lower extremity running kinematics between males and females in classified as various groups (normal/healthy runners, excessive pronators, rearfoot and forefoot strikers). With an expansive reference bank of three dimensional kinematics these same groups can be studied for comparison of various footwear conditions (shod, soft orthotic, semi-rigid orthotic, hard orthotic). This data could likely aide functional understanding of normal and abnormal gait mechanics as a result of pain or pathology. Evaluation of the presence of variability using continuous methods (dynamical systems) is better suited to this process. Changes in movement patterns and movement variability in real time can be noted and discussed for interpretation within the context of "optimal" movement variability.

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