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Effects of Orthotic Intervention during Running among Individuals with Functional Flatfoot

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EFFECTS OF ORTHOTIC INTERVENTION DURING RUNNING AMONG INDIVIDUALS WITH FUNCTIONAL FLATFOOT

by

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ABSTRACT

Foot orthotics are commonly prescribed to runners with functional flatfoot (FFF) with the goal of restoring the medial arch of the foot. In addition, treadmills are typically used by both clinicians and researchers in order to measure the lower extremity kinematics associated with running. However the mechanism of orthotic intervention as well as the accuracy of treadmills in representing overground running remains controversial within the literature.

This thesis first compared the lower extremity kinematics between treadmill and overground running among individuals with a subtalar neutral foot type. The results indicated no significant differences with respect to rate of rearfoot angle, maximum internal tibial rotation angle and rate of internal tibial rotation between the two running surfaces. However, maximum rearfoot angle was significantly higher during treadmill running. In addition, this thesis compared the lower extremity kinematics during running between individuals with subtalar neutral and FFF foot types. The results indicated similar lower extremity kinematics during running between groups as no significant differences were found between maximum rearfoot angle, rate of rearfoot angle or rate of internal tibial rotation. However, the subtalar neutral group demonstrated significantly higher maximum internal tibial rotation angles when compared to the FFF group. Finally, this thesis investigated the effects of orthotic intervention on the lower extremity kinematics during running among individuals with FFF. The results suggest that orthotics significantly decrease maximum rearfoot angle and maximum internal tibial rotation angle during running among this population. However, rate of rearfoot angle and rate of internal tibial rotation were not affected.

These findings suggest that treadmills do accurately represent the lower extremity kinematics associated with overground running, however if clinical decisions are dependent on small changes in maximum rearfoot angle then careful interpretation should be employed when using treadmills. Individuals with FFF did not demonstrate the expected increase in lower extremity kinematics therefore further research is required to better understand the mechanism of running injury among this population. In addition, orthotic intervention may have a mechanical effect on the motion of the lower extremity however the extent and applicability of this effect should be further examined.

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The thought of even completing a Masters degree would have been inconceivable had it not been for a number of individuals.

I'd like to thank my supervisor, Stephen Perry, for making my Masters experience such a positive one. You challenged me yet provided the perfect amount of guidance to optimize my learning. Thank you for sharing your knowledge, passion and creativity with me. I truly feel that your mentoring has enabled me to grow both personally and professionally. For this I am truly grateful.

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To all of my friends who now use the terms gait and pronation in their everyday vocabulary...thank you, thank you, and thank you! I am so fortunate to have you all in my life.

Finally, and perhaps most importantly, I'd like to thank my family for all of the love and support I have felt throughout my life. Particularly to my parents, Glen and Mary Lou, your active involvement and excitement in everything I do is unquestionably the greatest gift. I would not have made it through this current endeavour without your constant encouragement and understanding. Thank you for always believing in me.

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Chapter 1: Introduction

1.1 Abstract

The popularity of running as a recreational activity has grown immensely over the past few decades (McClay, 2000). It has been estimated that there are approximately 40 million recreational runners in North America (McKenzie, Clement and Taunton, 1985). Consequently, running related injuries are commonly seen in rehabilitation clinics. Studies indicate that 60-65% of all runners are injured during an average year which is defined as a physical problem severe enough to force a reduction in training. In addition, runners have been shown to miss approximately 5-10% of their scheduled training due to injury whereas racewalkers miss just over 1% (Byrnes, McCullagh, Dickinson and Noble, 1993). These injuries are associated with every foot type, however, there appears to be a higher incidence of running related injuries among individuals who demonstrate abnormal running mechanics (Subotnick, 1985; McClay and Manal, 1998). Individuals with functional flatfoot (FFF) have been shown to excessively and abnormally pronate during the stance phase of running thus, increasing their risk of developing an injury to the foot (McClay and Manal, 1998; Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005). Through the theory of joint coupling, this excessive foot motion may also result in excessive torsional motion of the tibia, thereby resulting in an increase in knee injuries. However, these results have not been consistent within the literature. As a result, there has developed a need to better understand the mechanism of running related injuries, particularly among individuals that demonstrate abnormal running mechanics as seen in FFF.

Currently, orthotic prescription is recommended for individuals with symptomatic FFF in order to control the excessive motion of the lower extremity during running. There is general agreement in the literature with respect to the clinical effectiveness of orthotic intervention among runners. In particular, the use of foot orthotics has been positively associated with patient satisfaction (Donatelli, Hurlbert, Conaway and St.Pierre, 1988; Moraros and Hodge, 1993) and pain reduction (Gross, Davlin and Evanski, 1991; Moraros and Hodge, 1993; Nawoczenski, Cook and Saltzman, 1995; Walter, Ng and Stoltz, 2004) thus enabling individuals to return to running (Donatelli, Hurlbert, Conaway and St.Pierre, 1988). Currently, researchers are attempting to understand the mechanism by which orthotics function in order to produce these encouraging symptomatic reductions. It has been speculated that orthotics may realign the lower extremity in order to decrease the excessive motion of the rearfoot and tibia that is typically seen among individuals with FFF (Mundermann, Nigg, Humble and Stefanyshyn, 2003; Nester, van der Linden and Bowker, 2003; MacLean, McClay Davis and Hamill, 2006). It seems that for every study that indicates a positive mechanical effect of orthotics in reducing excessive motion of the lower extremity (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Mundermann, Nigg, Humble and Stefanyshyn, 2003; Nester, van der Linden and Bowker, 2003; MacLean, McClay Davis and Hamill, 2006) there is a study reporting that orthotic intervention has no mechanical effect on the lower extremity (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Kitaoka, Luo, Kura and An, 2002; Williams III, McClay Davis and Baitch, 2003; Stackhouse, McClay Davis and Hamill, 2004).

The following sections within this chapter will provide a more in depth analysis of the current literature as it pertains to running related injuries (1.2), running mechanics of FFF (1.3) and the effectiveness of orthotic intervention (1.4). Section 1.5 will discuss the accuracy of kinematic measurements including marker type and location as well as running surface. The research goals and hypotheses will be presented in section 1.6 followed by the references in section 1.7.

1.2 General introduction to running as a recreational activity

There has been a large increase in the number of recreational runners over the past few decades (McClay, 2000). For example, the Boston Marathon, which began in 1897 with a total of 18 entrants, now sees over 20,000 participants each year (Boston Athletic Association, 2008). As a result, more and more individuals are reaping the health benefits that are associated with an increase in cardiovascular fitness as a result of running. In addition to the physical health benefits that are associated with running, researchers have shown a link between cardiovascular fitness and mental health. This link demonstrates that an increase in cardiovascular fitness is associated with lower depressive symptomatology (Galper, Trivedi, Barlow, Dunn and Kampert, 2006; Smith, Blumenthal, Babyak, Georgiades, Hinderliter and Sherwood, 2007) and a greater emotional well-being (Galper, Trivedi, Barlow, Dunn and Kampert, 2006). Due to the positive health benefits associated with running, as well as the relative affordability of this activity, it is not surprising that running has developed into one of the most popular fitness activities.

1.2.1 Mechanics of running

Interest into the underlying mechanics of running was ignited by the exponential growth of running as a recreational activity during the 1960's and 1970's (McClay, 2000). Early researchers in this field divided running into three phases: stance, swing and flight phases. This thesis will focus on the stance phase, which occurs when the foot is in contact with the ground. The stance phase can be further divided into sub-phases which contain the events of heel contact, midstance and toe-off. Figure 1.1 depicts the motion of the lower extremity during one complete gait cycle. There are both kinematic and kinetic changes that occur at each portion of the stance phase during running.

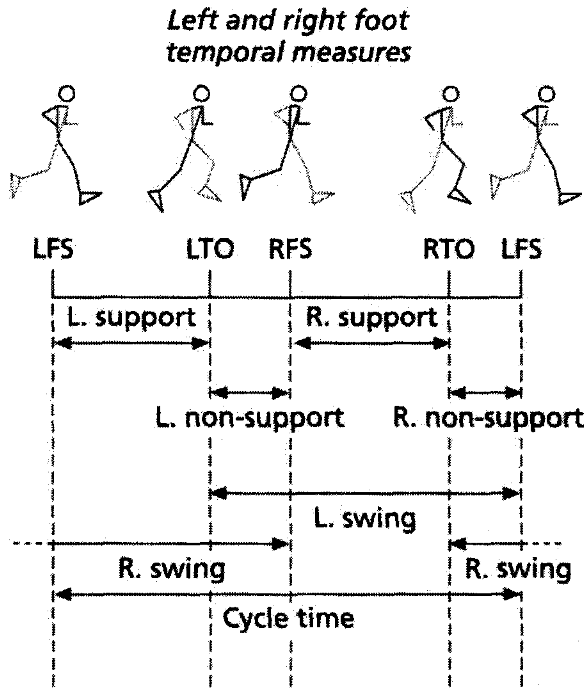

Figure 1.1: Lower extremity temporal measures during running (LFS=left foot strike; LTO=left toe off; RFS=right foot strike; RTO=right toe off) (Williams, 2000).

1.2.1.1 Kinematic changes during the stance phase of running

Kinematic investigation into the stance phase of running involves understanding the motion of the lower extremity as it is in contact with the ground. At heel contact, the lateral aspect of the calcaneus typically strikes the ground with the foot in an inverted, dorsiflexed position and the knee slightly flexed. The foot then begins to pronate and plantarflex at the subtalar joint while the leg begins to internally rotate in order to function as a mobile adapter and ultimately absorb shock. In the next phase, midstance, the foot moves towards a supinated position in order to lock the subtalar joint. At the same time, the knee extends and the leg externally rotates. These movements at the foot and knee act to move the body from a mobile adapter to a rigid lever in order to support the bodyweight as it passes over the leg and foot and prepares for the final stance phase of toe off. At toe off the ankle reaches its maximal plantarflexed position as the heel lifts from the ground, the foot is in a supinated position and the knee is flexed in order to propel the body forward (Perry, 1992; Whittle, 1999).

1.2.1.2 Kinetic changes during the stance phase of running

When examining the kinetics or forces that occur during running, researchers are typically interested in the vertical (Fz) and anterior-posterior (Fy) forces that are produced when an individual steps onto a force plate (Figure 1.2). In the Fz direction, the heel strikes the force plate which results in an initial impact peak. As the foot moves towards midstance, the body continues to accelerate downwards and the muscles of the lower extremity begin to contract in order to accept and support the body weight. These muscles continue to fire when moving into the toe off phase in order to propel the body forward. As a result, the vertical force during running contains one passive peak (due to the initial impact) and one active peak (due to the contraction of the musculature). The anterior-posterior force (Fy) during running contains both braking and propulsive forces. At heel contact, force is applied by the foot to the floor in an anterior direction in order to slow the forward momentum of the body. During midstance, the body is directly over the foot which results in no anterior-posterior force being applied. As the foot moves towards toe off, force is applied by the foot to the floor in a posterior direction in order to propel the body forward.

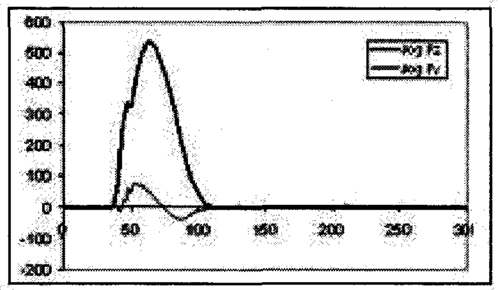

Figure 1.2: Vertical (Fz) and anterior-posterior (Fy) forces during the stance phase of running (horizontal axis: time (ms); vertical axis: force (N)).

1.2.2 Theory of lower extremity joint coupling

As knowledge concerning the mechanics of running progressed, researchers began to examine the coordination of motion between joints and segments of the lower extremity. Thus, the theory of joint coupling was developed in an attempt to explain the movement of the lower extremity during running. This theory implies that movement and forces occurring at the foot are transferred to the tibia during running. Figure 1.3 depicts the anatomy of the foot which will help in understanding this theory.

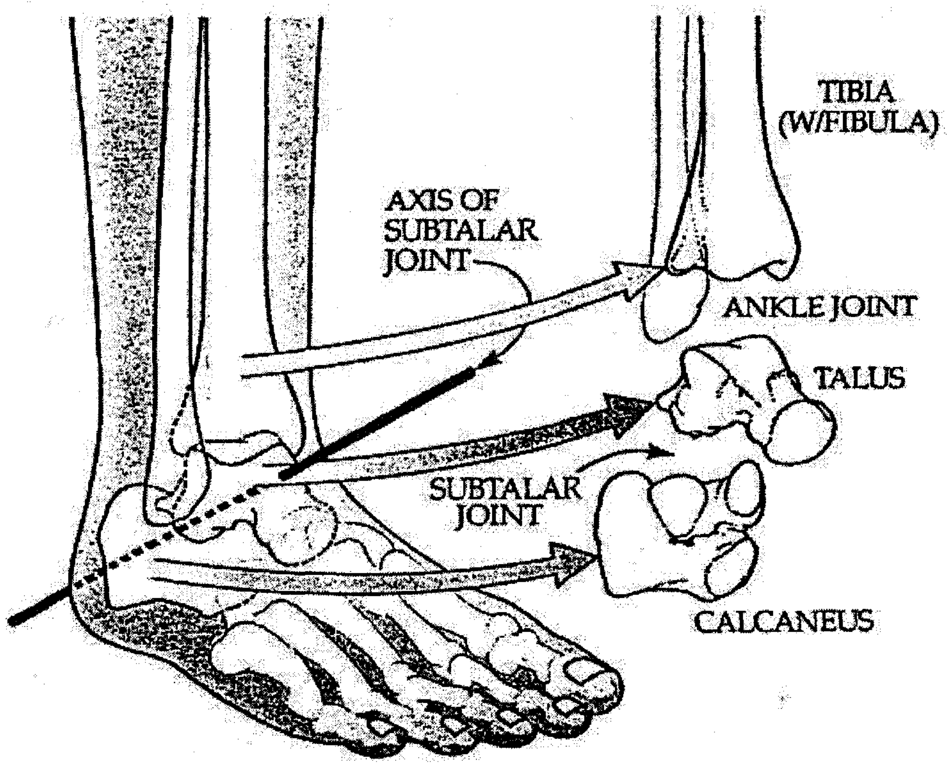

Figure 1.3: Anatomy of the lower extremity (NIKE, 1989).

During running, the initial movement of the foot at heel contact is pronation. Foot pronation is a complicated, tri-planar movement as it consists of eversion, abduction and dorsiflexion of the calcaneus relative to the talus. During the stance phase of running, the calcaneus is unable to abduct as it is in contact with the ground. As a result, the talus adducts on the calcaneus in order to achieve pronation and attenuate force. The movement of the talus causes the tibia to rotate internally due to the tight articulation of this joint (DeLeo, Dierks, Ferber and Davis, 2004). Previous research has provided evidence for the theory of joint coupling by demonstrating an association between calcaneal and tibial movement during each portion of the stance phase while running.

More specifically, research has shown that from heel contact to midstance the calcaneus everts and the tibia rotates internally and from midstance to toe off the calcaneus inverts and the tibia rotates externally (Stacoff, Nigg, Reinschmidt and van den Bogert, 2000; Eslami, Begon, Farahpour and Allard, 2007). Further support for the lower extremity joint coupling theory has come from the early studies of relative timing. These studies suggest that there is a synchrony between peak eversion, peak internal tibial rotation and peak knee flexion which occurs near midstance (McClay and Manal, 1997).

The difficulty in measuring the orientation of the subtalar joint (the articulation between the talus and the calcaneus) without using invasive techniques has lead researchers to develop the rearfoot eversion to tibial internal rotation (EV/TIR) ratio. This ratio provides a measure of the relative motion between eversion and tibial internal rotation from heel contact to the respective peaks occurring around midstance which is suggestive of the subtalar joint orientation (DeLeo, Dierks, Ferber and Davis, 2004). Typically, EV/TIR ratios during running are between 1 and 2 which indicates there is greater rearfoot eversion when compared to tibial internal rotation (Stacoff, Nigg, Reinschmidt and van den Bogert, 2000; Eslami, Begon, Farahpour and Allard, 2007). For example, an EV/TIR ratio of 2 indicates that every 2 deg of eversion will result in 1 deg of tibial internal rotation. Thus, the development of the EV/TIR ratio has enabled researchers to non-invasively measure the orientation of the subtalar joint, with many studies providing further support for the theory of lower extremity joint coupling.

The controversy surrounding the theory of joint coupling developed once researchers attempted to understand how the medial longitudinal arch structure of the foot related to the EV/TIR ratio. Many studies have examined this relationship with the results lacking systematic agreement (Nigg, Cole and Nachbauer, 1993; McClay and Manal, 1997; Williams III, McClay, Hamill and Buchanan, 2001). Nigg et al. (Nigg, Cole and Nachbauer, 1993) examined the effects of arch height of the foot on angular motion of the lower extremity during running. With respect to eversion movement, they found no correlation between this variable and arch height (r^2=0.059, p<0.197). A significant correlation was found between arch height and maximal internal leg rotation such that individuals with high arches demonstrated greater maximal internal leg rotation (r^2=0.152, p<0.033). However, the 95% confidence limit for the estimation of the regression line was essentially horizontal indicating that this influence may be negligible. Thus, a lower EV/TIR ratio was seen in the high arch when compared to low arch runners. Further, a significant correlation was found between transfer coefficient and arch height (r^2=0.267, p<0.0034). More specifically, an increase in the transfer of foot eversion to internal leg rotation was seen with increasing arch height. Williams et al. (Williams III, McClay, Hamill and Buchanan, 2001) also observed a lower EV/TIR ratio among high arched runners when compared to low arched runners. Contrary to the results of Nigg et al. (Nigg, Cole and Nachbauer, 1993), both groups in this study showed similar tibial internal rotation excursions. Therefore, the difference in EV/TIR ratio was a result of a significantly higher rearfoot eversion excursion among the low arched runners. However, McClay and Manal (McClay and Manal, 1997) found that excessive pronators demonstrated a significantly lower EV/TIR ratio when compared to individuals with normal rearfoot mechanics. Although this study did not evaluate arch structure, they reported that the lower EV/TIR ratio was a result of an increase in tibial internal rotation among the pronator group since there were similar rearfoot eversion excursions across both groups. Therefore, it appears that a relationship may exist between arch structure and EV/TIR ratios however, the extent of this relationship remains unclear in the literature.

With respect to injury, researchers hypothesized that runners with lower EV/TIR ratios (more tibial rotation) would experience an increase in knee injuries and that runners with higher EV/TIR ratios (more eversion movement) would experience an increase in foot injuries (McClay and Manal, 1997; Williams III, McClay, Hamill and Buchanan, 2001). However, Williams et al. (Williams III, McClay, Hamill and Buchanan, 2001) discovered that the opposite was occurring. Individuals with low arches (higher EV/TIR ratios) and those with high arches (lower EV/TIR ratios) were experiencing an increase in knee injuries and foot injuries, respectively. Although Nigg et al. (Nigg, Cole and Nachbauer, 1993) found a correlation between the transfer coeffient and arch height, only 27% of the variance was explained by arch height. This suggests that other factors may be involved in determining the transfer of foot eversion to internal tibial rotation in order to explain the mechanism of running-related injuries.

1.2.3 Running-related injuries

There is a higher prevalence of injury that occurs during running when compared to other activities, particularly walking. Researchers have identified many differences between the mechanics of walking and running during the stance phase which may explain the difference in injury rates. These differences are outlined below.

Walking is comprised of three phases: single support, double support and a swing phase. The double support phase in walking occurs when both feet are in contact with the ground and allows for a greater distribution of force. Running is comprised of two phases: a single support and a flight phase (which occurs when the body is airborne). Since there is no double support phase during running, the forces are transferred to the foot and lower extremity during the single support phase. The typical stance phase while running lasts 0.25s compared to 0.8s while walking with rapid pronation occurring over 0.015s to 0.030s. Consequently, rapid pronation while running occurs in one-fifth the time that would typically be seen while walking (Subotnick, 1985). Therefore, the forces are distributed over a smaller area and at a faster rate during running, which may increase the amount of force that is transmitted to a specific location on the foot. If a specific region of the foot is experiencing an increase in force, this may predispose the runner to injury.

A difference in the amount of force applied to the body has been observed between walking and running. The ground reaction force (GRF) that is applied to the body during walking is typically equivalent to the individual's bodyweight. However, the GRF that is transferred to the body during running is three to five times the individual's bodyweight (Subotnick, 1985). One study found that the resultant GRF's increased from walking (peak 0.51 BW and loading rate 24.6 BW/s) to running (1.96 BW and 115.2 BW/s) (Perry and Lafortune, 1995). The combination of an increase in GRF occurring at a faster rate results in less time to attenuate a greater amount of shock. As such, these factors may predispose the body to a higher risk for injury during running than walking.

Kinematic differences have also been noted between walking and running. A study conducted by Perry and Lafortune (Perry and Lafortune, 1995) found the maximum rearfoot angle was 4.4 deg greater during running than walking. Further, it has been suggested that excessive foot eversion may cause excessive tibial internal rotation by means of joint coupling at the ankle (Clement, Taunton, Smart and McNicol, 1981). Thus, running injuries may also occur at the knee as well as at the foot. The 'rule of three' suggests that biomechanical imbalances in the lower extremity are approximately three times more important to the runner than the walker (Subotnick, 1985). For example, a 4 deg imbalance of the rearfoot in a walker would be as significant as a 12 deg imbalance if that walker began running. Therefore, even a slight increase in rearfoot eversion may result in injury during running which otherwise may not have developed if the individual was only walking.

Most running injuries occur due to overuse or as a result of microtrauma which are the stresses that are absorbed during distance running. For the reasons presented above, runners are at a higher risk of developing an injury when compared to walkers. It has been proposed that almost perfect biomechanics would be required in order to successfully run long distances on artificial surfaces (Subotnick, 1985). Therefore, it is no surprise that runners who present with abnormal biomechanics may be at an even higher risk of developing injury.

1.3 Functional flatfoot as a pathological condition

The condition of flatfoot is characterized by a significant reduction in the medial longitudinal arch of the foot. A flatfoot is typically categorized as either functional or rigid. The term functional flatfoot (FFF) refers to the high degree of flexibility within the foot during physical examination when compared to a rigid flatfoot. Therefore, a FFF differs from a typical flatfoot in that the reduction of the arch is seen while weight bearing. For example, an arch exists under the medial aspect of the foot while sitting (Figure 1.4) but disappears completely or is significantly reduced with weight bearing (Figure 1.5) (Kim and Weinstein, 2000). The focus of this thesis will be on individuals with FFF.

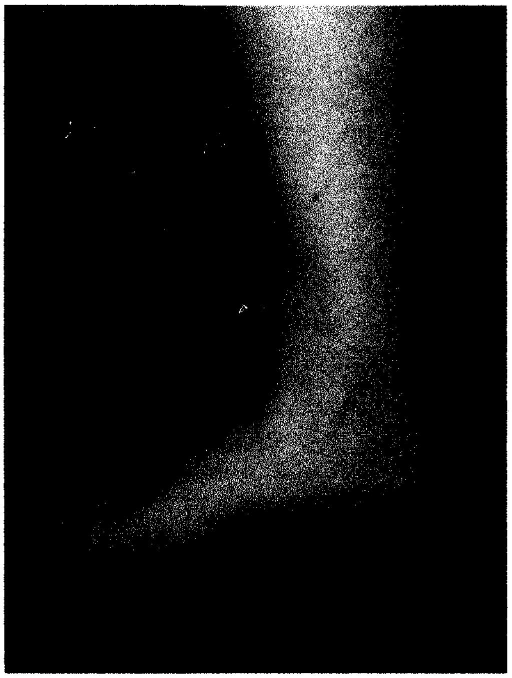

Figure 1.4: FFF while non-weight bearing

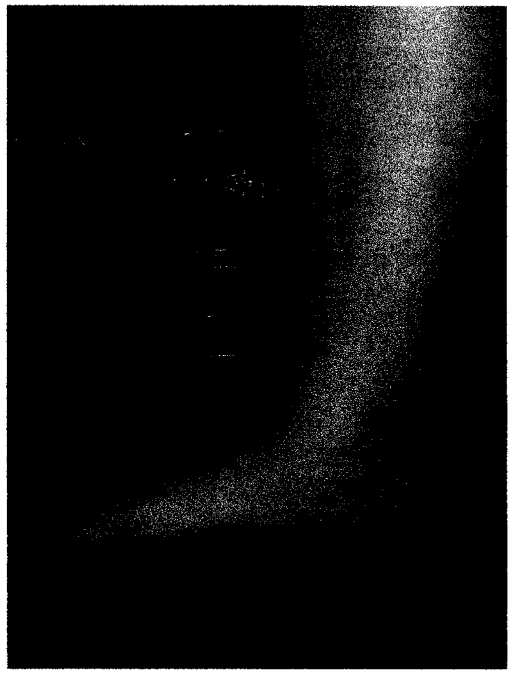

Figure 1.5: FFF while weight bearing

The development of a medial longitudinal arch is a natural process of childhood growth. It has been suggested that 90% of infants and toddlers up to the age of two years have varying degrees of flatfeet due to the presence of an infant fat pad along the medial aspect of the foot and the normal joint hypermobility typically seen during this age group (Staheli, Chew and Corbett, 1987). However, most children with flatfeet do not present with any degree of discomfort or disability (Kim and Weinstein, 2000). The medial longitudinal arch is thought to develop between three and five years of age in most children with only 4% of children demonstrating flatfeet by the age of 10 (Staheli, Chew and Corbett, 1987).

Adult FFF is generally thought to be a progression of the condition from childhood as it is also characterized by partial or complete loss of the medial longitudinal arch while weight bearing (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005). FFF deformity is frequently encountered within the adult population and may present with clinical consequences ranging from mild limitations to severe disability and pain (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005). Although the exact cause of FFF is unknown, it is thought that family history, activity level, obesity, footwear and occupation may be some contributing factors (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005).

1.3.1 Anatomy of the lower extremity in functional flatfoot

The reduction in the medial longitudinal arch has been thought to result in abnormal and excessive pronation of the foot. Figure 1.6 illustrates the structures and joints that are involved with pronation. During pronation, the talus adducts and plantarflexes on the calcaneus, which simultaneously everts and plantarflexes (Donatelli, 1997). The movement of the talus causes the tibia to rotate internally due to the tight articulation of the talocrural joint.

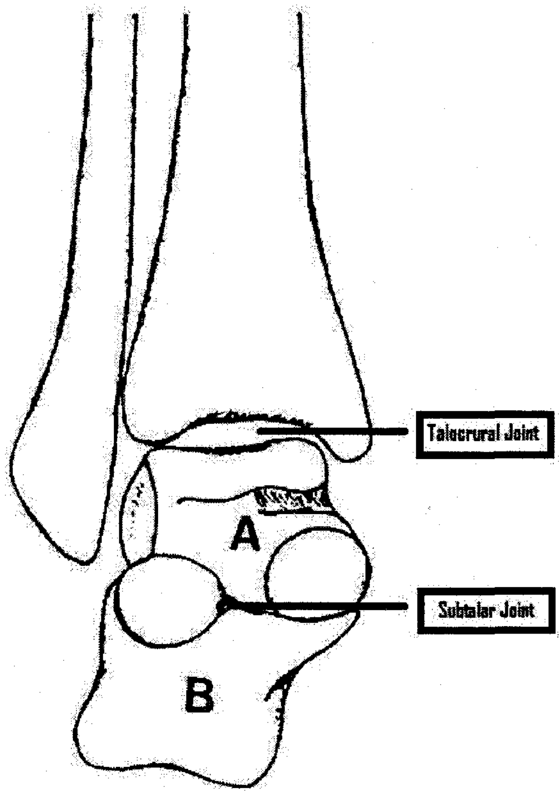

Figure 1.6: Anterior view of the closed kinetic chain pronation. The calcaneus (B) everts and the talus (A) adducts and plantarflexes (Donatelli, 1997).

Pronation of the foot can also been seen when looking at the posterior aspect of the tibiocalcaneal alignment. Figure 1.7 demonstrates the rearfoot tibiocalcaneal alignment that is seen among individuals with different foot types. A neutral foot type results in a neutral calcaneal position when compared to the tibia. While a supinated and pronated foot result in calcaneal inversion and eversion, respectively, when compared to the tibia.

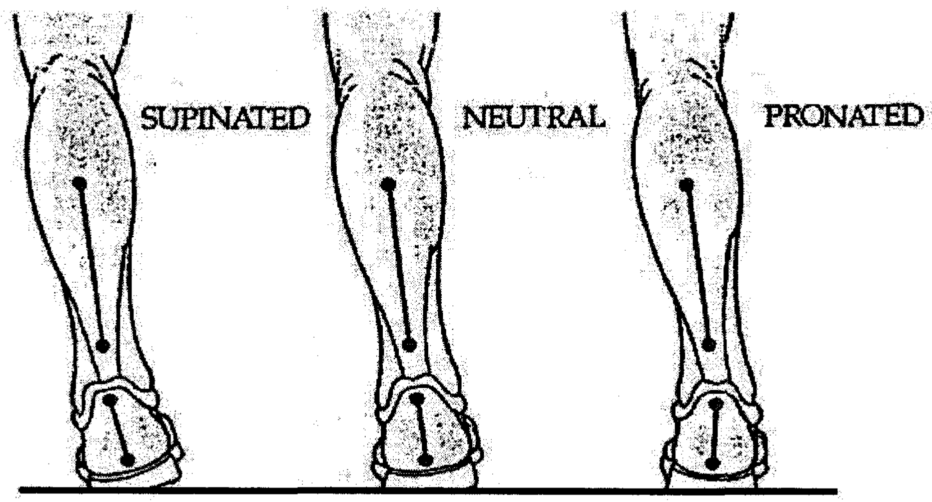

Figure 1.7: Tibiocalcaneal angles for 3 different foot types. A supinated and pronated foot type results in calcaneal inversion and eversion respectively when compared to the tibia (Nike, 1989).

1.3.2 Mechanics of running among the functional flatfoot population

Researchers are beginning to examine the differences between foot types with respect to running mechanics. More specifically, studies have demonstrated that lower extremity mechanics during running differ among the FFF and 'normal' populations (McClay and Manal, 1997; McClay and Manal, 1998; Hunt and Smith, 2004). McClay and Manal (McClay and Manal, 1997) produced one of the first studies which compared the coupling parameters in runners with normal (NL) and excessive (PR) pronation. The results indicate higher peak rearfoot eversion and tibial internal rotation excursions (tibia relative to the calcaneus) among the PR group. In addition, the PR group demonstrated a significantly lower EV/TIR ratio. Surprisingly, both groups demonstrated similar eversion excursions however the PR group demonstrated significantly higher tibial internal rotation excursions which resulted in the lower EV/TIR ratio.

In another study, McClay and Manal (McClay and Manal, 1998) examined the lower extremity kinematics between excessive pronators (PR) and normal pronators (NL) during running. The results showed significant kinematic differences between these two populations. They found that the foot was more everted at heel contact and toe off within the PR group when compared to the NL group (HC: PR=-8.5 deg, NL=1.7 deg; TO: PR=-4.8 deg, NL=1.7 deg). Further, the magnitude of rearfoot eversion among the PR group (-21.2 deg) was almost twice that of the NL group (-11.2 deg). The mean peak velocity of foot eversion was significantly greater among the PR group. However, no significant differences were found with respect to knee internal rotation of the tibia relative to the femur or knee internal rotation excursions between the PR and NL groups.

Although it is generally believed that different running mechanics occur between excessive and normal pronators, there has been some evidence to suggest that this may not be the case. Research conducted by Hunt and Smith (Hunt and Smith, 2004) examined the mechanics of the symptomatic flatfoot verses the normal foot during walking. They hypothesized that there would be an increase in rearfoot (frontal plane) and forefoot (sagittal plane) motion among the flatfoot group. In contrast to previous studies they reported no significant differences between frontal plane rearfoot motion (flatfoot, 9 deg; normal foot, 8 deg) and sagittal plane forefoot motion (flatfoot, 10 deg; normal foot, 12 deg) between these two groups. Despite finding some small significant differences between the groups with respect to foot motion they concluded that there was more of a restraint of motion while walking among symptomatic flatfoot individuals when compared to a normal foot. Thus, further research may be required to better understand the mechanics that are occurring among the symptomatic FFF population during both walking and running.

The effects of arch height on lower extremity kinematics have also been investigated during running (Nigg, Cole and Nachbauer, 1993; Williams III, McClay, Hamill and Buchanan, 2001; Williams III, McClay Davis, Scholz, Hamill and Buchanan, 2004). As previously discussed in section 1.2, Nigg et al. (Nigg, Cole and Nachbauer, 1993) found no significant influence of arch height on either maximal eversion movement or maximal internal leg rotation during the stance phase of running. However, the transfer of foot eversion to internal leg rotation was significantly increased with an increased arch height, indicating that a functional relationship may exist between arch height and injury. Since arch height only accounted for 27% of this coupling, the authors concluded that there must be other contributing factors which should be further investigated.

Conversely, Williams III et al. (Williams III, McClay, Hamill and Buchanan, 2001) examined the effects of arch height on the lower extremity kinematics during running and found significant differences between high and low arched individuals. Low arched individuals demonstrated an increased rearfoot eversion excursion and rearfoot eversion velocity when compared to high-arched individuals. In addition, the increase in eversion excursion resulted in an increase in the EV/TIR ratio among low arched individuals.

A difference in the methodology of determining arch height exists within the literature. Nigg et al. (Nigg, Cole and Nachbauer, 1993) determined arch height as being the highest point along the medial plantar curvature while in a full weight bearing position with the foot resting lightly on a raised platform. Arch height was measured using a digital caliper. Other researchers (Williams III, McClay, Hamill and Buchanan, 2001; Williams III, McClay Davis, Scholz, Hamill and Buchanan, 2004) have determined the arch height ratio as the height of the medial arch from the floor at 50% of the foot length divided by the truncated foot length (TFL). TFL is defined as the distance from the posterior aspect of the calcaneus to the medial joint space of the first metatarsal phalangeal joint.

Arch height ratio = arch height / truncated foot length

These differences in arch height measurements may reflect the varying results seen between these studies. In order to accurately compare the effects of arch height on running mechanics, a standardized method for measuring arch height should be determined.

Lower extremity electromyography and kinetics have also been evaluated between low and high arched individuals. It has been suggested that low arched individuals have a decrease in leg stiffness (Williams III, McClay Davis, Scholz, Hamill and Buchanan, 2004), knee stiffness (Williams III, McClay Davis, Scholz, Hamill and Buchanan, 2004) and vertical loading rates (Williams III, McClay, Hamill and Buchanan, 2001; Williams III, McClay Davis, Scholz, Hamill and Buchanan, 2004) when compared to high arched individuals. These factors may further contribute to the risk of running related injuries among FFF individuals.

1.3.3 Running-related injuries among the functional flatfoot population

The different lower extremity mechanics among FFF individuals may be associated with an increased risk of developing a running related injury (Subotnick, 1985; McClay and Manal, 1997; McClay and Manal, 1998; Kaufman, Brodine, Shaffer, Johnson and Cullison, 1999; McClay, 2000). Previous literature suggests that a FFF excessively pronates for a longer period of time (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005) or even throughout the entire stance period (McClay and Manal, 1998) without ever moving towards a supinated position. The normal function of pronation is to unlock the subtalar joint in order to better attenuate the forces that are applied to the body during heel contact to midstance. Then typically, from midstance to toe off, the foot supinates in order to lock the subtalar joint, causing the foot to become a stable rigid lever. With the foot as a rigid lever, it is better able to support the weight of the body as it passes over the leg and foot during normal gait. In addition, a rigid lever allows for optimal and successful completion of toe off. The fact that a FFF may be pronated for the entirety of stance inhibits the foot from supinating during midstance and toe off. Therefore, the pronated foot is supporting the weight of the body during the entirety of stance while the subtalar joint is in a very unstable position, thereby increasing the risk of injury.

Previous research has shown a link between excessive or prolonged pronation and injury during running (Nike, 1989; McClay and Manal, 1998). McClay and Manal (McClay and Manal, 1998) reported that the incidence of running-related injuries was significantly higher among excessive pronators (67%) when compared to normal pronators (22%). Typical overuse running injuries were reported including patellar tendonitis, shin splints, Achilles tendonitis, knee ligament damage and ankle sprains. Other common injury sites among individuals with symptomatic FFF include pain at the arch, heel and lateral border of the foot which increases with weight bearing activities such as running (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005).

Knee injuries are reportedly the most common injury site among runners (Clement, Taunton, Smart and McNicol, 1981). As previously described, excessive pronation may result in an increase in internal tibial rotation during stance due to the tight articulation at the talocrural joint. Previous research has demonstrated an increase in tibial internal rotation excursion among excessive pronators which may place this group at an increased risk of developing knee injuries during running (McClay and Manal, 1997). Controversy regarding the relationship between excessive pronation and tibial stress fractures exists within the literature. Kaufman et al. (Kaufman, Brodine, Shaffer, Johnson and Cullison, 1999) demonstrated an increase in tibial stress fractures among excessive pronators whereas, Hetsroni et al. (Hetsroni, Finestone, Milgrom, Ben-Sira, Nyska, Mann, Almosnino and Ayalon, 2008) found the opposite was true.

A difference in the EV/TIR ratio may place runners at risk for different types of injury (Nigg, Cole and Nachbauer, 1993; McClay and Manal, 1997; Williams III, McClay, Hamill and Buchanan, 2001). It is generally thought that an increase in the EV/TIR ratio (increase in rearfoot eversion excursion) will result in foot and ankle injuries whereas a decrease in the EV/TIR ratio (increase in tibial internal rotation) will result in knee injuries. However, there has been limited research conducted to determine the relationship between EV/TIR ratio and injury. Williams III et al. (Williams III, McClay, Hamill and Buchanan, 2001) examined this relationship and surprisingly found that low arched runners with a higher EV/TIR ratio (more rearfoot eversion excursion) actually had an increase in knee injuries, while high arched runners (low EV/TIR ratio) experienced more foot and ankle injuries. Therefore, the EV/TIR ratio may be clinically relevant however currently it does not provide a logical sequence and thus, more research needs to be conducted in order to verify these findings. Although there is some discrepancy, the majority of the literature suggests that individuals with FFF are at an increased risk of developing a running related injury due to the excessive rate and angle of pronation.

1.4 Effectiveness of foot orthotics

In order to control the excessive motion that has been documented among runners with FFF, foot orthotics are commonly prescribed. From a mechanical perspective, the assumption is that foot orthotics may optimize the skeletal alignment of these individuals bringing the movement of the lower extremity to resemble a more normal pattern during running. As a result, foot orthotics have been shown to be effective in treating problems associated with the foot, ankle and skeletal alignment (Landorf and Keenan, 2000). Previous research on the effectiveness of orthotics has typically been categorized into two key areas: the clinical effectiveness of orthotics including symptom relief and the mechanical function of the lower extremity while wearing orthotics.

1.4.1 Clinical effectiveness

There is reasonably consistent agreement in the literature regarding patient satisfaction and symptom relief associated with wearing foot orthotics. Further, it has been documented that runners are able to return to their sport if orthotics have been worn following injury.

A retrospective survey of 81 patients conducted by Donatelli et al. (Donatelli, Hurlbert, Conaway and St.Pierre, 1988) found that 91% of surveyed patients were satisfied with their orthotics. Further, at the time of the survey, 94% of patients were still wearing their orthotics and 52% stated that they would not leave home without them. Moraros and Hodge (Moraros and Hodge, 1993) completed a nationwide survey on patient satisfaction with prescription foot orthotics. Individuals were evaluated during 14 weeks in order to make any modifications to their foot orthotics. Of the 403 respondents at the end of the 14 weeks, 83.1% of patients indicated that they were satisfied with their orthotics. Further, 70.5% stated that the final fit of the custom orthotic was excellent; with 29.3%, 1.7% and 1.0% reporting that it was good, fair and poor, respectively.

Foot orthoses have been successfully used to treat various lower extremity symptoms including, but not limited to, knee pain, plantar fasciitis, shinsplints, iliotibial band tendonitis (Nawoczenski, Cook and Saltzman, 1995) and mild to moderate osteoarthritis at the medial knee (Rubin and Menz, 2005). The results of the nationwide survey by Moraros and Hodge (Moraros and Hodge, 1993) on the effectiveness of orthotics to resolve the primary complaint indicate that at the final visit (14^{th} week), 62.5% had their chief complaint completely resolved; 32.8% partially resolved; and 4.7% unresolved. Walter et al. (Walter, Ng and Stoltz, 2004) completed a survey regarding symptomatic relief among patients who wore prescribed functional foot orthotics for a minimum of one year. Of the 266 respondents, 75% reported pain relief (ranging from 60%-100%) after wearing orthotics. However, 9% of patients stated that the orthotics did not help at all in reducing pain. In another survey of 500 long-distance runners who had been prescribed orthotic shoe inserts, 76% reported a complete or great improvement of their symptoms (Gross, Davlin and Evanski, 1991). Orthotic shoe inserts were most effective among individuals suffering from biomechanical abnormalities including excessive pronation. Further, 90% of the participants in this study continued to wear their orthotics even after their symptoms had disappeared.

In addition, studies have been conducted to compare the effectiveness of different types of foot orthotics in reducing pain. Rome et al. (Rome, Gray, Stewart, Hannant, Callaghan and Hubble, 2004) compared the clinical effectiveness of two types of foot orthotics commonly used to treat plantar heel pain. The orthotics used in this study were accommodative (used to provide cushioning, padding and shock absorption during gait) and functional (used to achieve weight bearing realignment of the lower extremity and to redistribute the load during gait). Participants were evaluated at baseline, after 4 weeks and after 8 weeks using The Foot Health Status Questionnaire (FHSQ) which determines foot health spanning 4 domains: foot pain, foot function, footwear and general foot health. The results showed a significant decrease in foot pain and a significant increase in foot function from baseline to 4 to 8 weeks with the functional orthotic as determined by the FHSQ. The accommodative orthotic demonstrated a significant decrease in foot pain from baseline to 4 weeks only, indicating that these orthotics reached their maximum potential to reduce pain at 4 weeks. The authors concluded that although the functional foot orthoses were initially more expensive, they resulted in a better quality of life.

In a study conducted by Donatelli et al. (Donatelli, Hurlbert, Conaway and St.Pierre, 1988), 70% of individuals who had sustained a running related injury were able to return to their sport with the use of a functional foot orthotic. Researchers are in relative agreement as to the effectiveness of orthotics in reducing pain associated with running related injuries. However, the mechanisms by which orthotics function to produce symptomatic relief remains inconclusive within the literature.

1.4.2 Mechanical function of the lower extremity

Although there has been considerable reported subjective relief while wearing orthotics, the scientific data related to understanding the mechanism through which orthotics function remains controversial in the literature. As previously noted, individuals with FFF are thought to have an increase in rearfoot eversion and subsequent internal tibial rotation, by means of joint coupling, while running. It has been speculated that orthotics may function by realigning the lower extremity in order to decrease the excessive motion of the rearfoot and tibia that is typically seem among these individuals.

Many studies have demonstrated a reduction in maximum rearfoot eversion during running while wearing a foot orthotic. MacLean et al. (MacLean, McClay Davis and Hamill, 2006) investigated the effects of custom foot orthotics on the lower extremity of 15 female runners. Participants completed 5 overground running trials, with and without a foot orthotic, while in a running shoe that had the heel counter removed. The results indicated a significant reduction in maximum rearfoot eversion and maximum rearfoot eversion velocity while wearing the custom foot orthotic. In a similar study, Mundermann et al. (Mundermann, Nigg, Humble and Stefanyshyn, 2003) examined the effects of posting and custom-molding of foot orthotics on lower extremity kinematics during running. Twenty one pronators were recruited for this study and wore running sandals during testing. The insoles of the running sandals were removed and replaced in order to complete the orthotic condition trials. The results indicated that medial posting within foot orthotics significantly decreased maximum foot eversion and maximum foot eversion velocity while running. Nester et al. (Nester, van der Linden and Bowker, 2003) investigated the effects of medially and laterally wedged foot orthotics on the kinematics of normal walking gait. They found that a 10 deg wedge within the medial aspect of the foot orthotic significantly reduced foot pronation. Earlier studies have examined the effects of orthotics on maximum foot eversion during running and have found similar results (Rodgers and Leveau, 1982; Smith, Clarke, Hamill and Santopietro, 1986).

Although a lot of convincing evidence exists for the use of orthotics in reducing maximum foot eversion, there also appears to be an equal amount of evidence that challenges this notion. In a cadaveric study, Kitaoka et al. (Kitaoka, Luo, Kura and An, 2002) assessed the effects of foot orthoses on arch height and rearfoot alignment. They found that arch height, compared with that of the flat foot condition, increased significantly but to a limited degree (<2%) while using a foot orthotic. However, the rearfoot alignment, as determined by the calcaneal-tibial position, did not improve with foot orthotics. Stackhouse et al. (Stackhouse, McClay Davis and Hamill, 2004) compared the effects of custom orthotics on forefoot and rearfoot strike running patterns. Although there were no significant findings for peak eversion, eversion velocity or eversion excursion with either foot strike, there were some subjects who demonstrated a decrease in these variables with orthotic use. This finding indicates that orthotic intervention may be highly individualized. Stacoff et al. (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000) examined the effects of medial foot orthoses on skeletal movements of the calcaneus and tibia during running. Their methodology differed slightly in that they inserted intracortical Hofman pins with reflective marker triads into the calcaneus and tibia in order to track the movement of those segments. The results of 5 participants indicated that medial orthotics reduced maximum eversion (except for one subject) however, this reduction was not significant. Similar findings have been found in other studies indicating that orthotic intervention may not have a significant effect on maximum foot eversion while running (Nawoczenski, Cook and Saltzman, 1995; Williams III, McClay Davis and Baitch, 2003).

More recently, studies have begun to investigate the association between symptomatic relief from orthotics and kinematic variables. Zammit and Payne (Zammit and Payne, 2007) conducted a study to examine the effects of foot orthoses on rearfoot motion and how these changes correlated with the degree of symptom relief. The results of 22 excessive pronators indicated that orthotics had a small but statistically significant effect on rearfoot motion. Using The Foot Health Status Questionnaire, pain and functioning levels were recorded at baseline and then again 4 weeks later. Interestingly, at the 4 week follow-up appointment they did not find a correlation between differences in rearfoot motion and positive clinical outcomes. Therefore, they concluded that the rearfoot motion changes were not accounting for the extent of symptom reduction seen in this study and that other mechanisms of orthotic function should be examined.

In terms of internal tibial rotation, again there is literature that both supports and refutes the effects of orthotic intervention on this kinematic variable. Bellchamber and van den Bogert (Bellchamber and van den Bogert, 2000) examined the cause and effect relationship between internal tibial rotation and pronation of the foot during heel to toe running. The goal of this study was to determine if orthotic intervention would be effective at reducing knee pain that was attributed to excessive internal tibial rotation. The results demonstrated periods during the stance phase of running where axial tibial rotation was driven by the foot. Thus, they proposed that orthotics would be effective at limiting axial tibial rotation during these time periods. As such, there have been a number of studies that have demonstrated a reduction in internal tibial rotation while using a foot orthotic.

Stacoff et al. (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000) investigated the effects of medial foot orthotics on skeletal movements of the calcaneus and tibia during the stance phase of running. A significant decrease in total internal tibial rotation was found as a result of the medial orthotic. Nawoczenski et al. (Nawoczenski, Cook and Saltzman, 1995) evaluated the effects of custom rigid orthotics among both low-arched and high-arched individuals. The use of orthotics produced a small but significant decrease in total range of tibial rotation relative to the rearfoot. However, it is interesting to note that both groups experienced a similar reduction in the tibial rotation range. A study conducted by Mundermann et al. (Mundermann, Nigg, Humble and Stefanyshyn, 2003) looked at the effects of posting, molding, and combined posting and molding within foot orthotics on the effects of lower extremity kinematics during running. Posting was found to significantly reduce maximum tibial rotation and maximum tibial rotation velocity. However, tibial rotation was affected to a lesser extent when compared to foot eversion movement. Both molding, and posting and molding were found to significantly reduce maximum tibial rotation.

Conversely, published research has shown that orthotic intervention has no effect on internal tibial rotation while running. Nigg et al. (Nigg, Khan, Fisher and Stefanyshyn, 1998) examined the effects of shoe inserts on total internal tibial rotation during running. Five shoe inserts had a bilayer design which was constructed using two different materials at the top and the bottom of the insert. These inserts were then tested against each other and also against a no shoe insert condition. The results indicate that the total internal tibial movement was slightly less during the insert conditions when compared to the no insert condition, however, these findings were not statistically significant.

There is a large amount of research to both support and challenge the mechanical effectiveness of foot orthotics during running. The current literature remains inconclusive as to whether orthotics effectively align the lower extremity during running in order to decrease excessive motion at the rearfoot and tibia. Much of the controversy in the literature can be attributed to differences in participants, orthotic construction, foot type, running speed and kinematic marker type and placement.

1.5 Accuracy of kinematic measurements

There are a number of factors to consider when attempting to measure lower extremity kinematic variables. The type and location of markers as well as footwear and running surface may all influence the outcome of the study. When comparing the results of previous studies surrounding running mechanics and orthotic intervention it is important to consider these differences in methodology as a possible explanation for the contradicting results currently seen within the literature.

1.5.1 Kinematic tracking markers

When measuring kinematic variables typically two types of markers are used: bone pins or skin markers. Bone pins are directly inserted into the bone to track movement. This option is fairly invasive as it requires a small surgery. As well, a large surgical triad projecting from the leg may affect the subject's typical gait mechanics. Another option would be to use skin markers which are placed directly on the skin in order to track the underlying bone movements. Proper location of skin markers is essential in order to obtain accurate kinematic measurements during gait. The closer the markers are to the bone the more accurate their measurements will be. Reinschmidt et al. (Reinschmidt, van den Bogert, Lundberg, Nigg, Murphy, Stacoff and Stano, 1997) investigated the errors at the tibiofemoral and tibiocalcaneal joints associated with skin movement artifact (skin markers). The results suggest that knee rotations may be affected with substantial errors when using skin markers however tibiocalcaneal movements are generally well reflected when using skin markers.

Skin markers were chosen for this research project and appropriate care was taken to place the external skin markers at locations with minimal adipose tissue and muscle between the bone and the skin to limit motion artifact. As such, bony landmarks including condyles and malleoli were used to track the movement of the lower extremity.

1.5.2 Use of footwear

Another factor to consider when measuring lower extremity kinematic variables is whether the participants will perform running trials while wearing shoes. Many studies have investigated the movement of the foot within the shoe during running and have found that footwear may influence the normal movement of the foot. Stacoff et al. (Stacoff, Reinschmidt and Stussi, 1992) measured the movement of the heel counter as well as the movement of the heel inside the shoe. They determined that the heel inside the shoe moves less and at a slower rate than the heel counter. Nigg (Nigg, 1986) demonstrated that a difference of 2-3 deg for the rearfoot markers can be observed between the heel counter and the heel. As a result, they propose that if footwear is to be worn then windows in the heel counter should be cut so that markers can be directly placed on the heel. However, cutting the heel counter may affect the structural integrity of the footwear which may negatively impact the results.

Studies involving torsion of the foot have been conducted to measure the effects of footwear of gait. Stacoff et al. (Stacoff, Kaelin, Stuessi and Segesser, 1989) examined the relationship between torsion and pronation in running during both rearfoot and forefoot strike patterns. The results indicated that compared to running barefoot, torsion of the foot was significantly restricted during running while wearing a shoe. This restriction in torsion resulted in a significant increase in foot pronation while wearing a shoe during both rearfoot and forefoot strike patterns. In another study involving torsion of the foot during running, Stacoff et al. (Stacoff, Kalin and Stussi, 1991) demonstrated that running shoes resulted in the greatest reduction of torsion angle which produced the largest increase in pronation angle, compared to running while wearing spikes or while barefoot.

Due to the possible confounding results with the use of footwear, the studies presented in this thesis were conducted barefoot in order to better control for the variables of interest.

1.5.3 Running surface

Most studies measuring gait mechanics perform running trials on either a treadmill or on an overground runway. The type of running surface that is used in the study may affect the results. Particularly the extent to which treadmills accurately represent overground locomotion remains controversial in the literature. Studies have shown that there are advantages and disadvantages associated with using this type of instrument to measure gait. Advantages of using treadmills in research include easy control of environmental variables such as velocity and incline, small testing area is needed and multiple gait trials can be recorded in less time when compared to overground runway trials. However, there are some disadvantages associated with using treadmills to measure gait. Disadvantages include difficulty measuring forces, the moving treadmill belt may interfere with natural gait mechanics and a familiarization period to treadmill running may exist in order for inexperienced treadmill runners to feel comfortable on the treadmill and to run as they typically would overground. In order to make comparisons between treadmill and overground studies the assumption is made that treadmills do not alter typical gait mechanics. Several studies have been conducted in order to test the accuracy of treadmill measurements. The results have been controversial with some studies supporting and other studies challenging the above assumption.

Schache et al. (Schache, Blanch, Rath, Wrigley, Starr and Bennell, 2001) measured the time-distance parameters and three-dimensional angular kinematics of the lumbo-pelvic-hip complex during treadmill and overground running. Significant differences were found for all of the time-distance parameters and lumbar extension and anterior pelvic tilt at initial contact and the first maximum anterior pelvic tilt. However, these significant results were not systematic across all subjects. Therefore Schache et al. (Schache, Blanch, Rath, Wrigley, Starr and Bennell, 2001) conclude that high powered treadmills with minimal belt speed fluctuations are capable of obtaining three-dimensional kinematic patterns of the lumbo-pelvic-hip complex during running. Van Ingen Schenau (Ingen Schenau, 1980) demonstrated through theoretical mathematical calculations that the use of a fixed coordinate system may lead to faulty conclusions regarding the description of treadmill locomotion. He suggested that as long as the belt speed is held constant a moving coordinate system should be used when analyzing treadmill data. He concluded that when this approach is used no mechanical differences were found between treadmill and overground locomotion. A study conducted by Frishberg (Frishberg, 1983) compared kinematic and temporal differences between overground and treadmill sprinting. Despite using a moving coordinate system as suggested by Van Ingen Schenau (Ingen Schenau, 1980), Frishberg (Frishberg, 1983) found significant kinematic differences between treadmill and overground running. Specifically, most of the kinematic differences occurred in or just before the support phase and concerned the supporting leg. With respect to the temporal components he reported no significant differences between the two modes of running, although each individual displayed definite trends. As well, Frishberg (Frishberg, 1983) found that oxygen debt was 36% higher in overground sprinting trials when compared to treadmill trials. Other studies have looked at treadmill familiarization as a necessary requirement in order to obtain similar measurements between treadmill and overground locomotion. Matsas et al. (Matsas, Taylor and McBurney, 2000) investigated knee kinematic differences and temporal and distance gait measurements between treadmill and overground walking. The results showed no significant differences in knee kinematics or temporal and distance gait measurements after running on a treadmill for 4 and 6 minutes respectively. A study by Lavcanska et al. (Lavcanska, Taylor and Schache, 2005) looked at the amount of time necessary for treadmill familiarization to occur among subjects with no prior treadmill experience. This study found significant differences at the pelvis, hip, knee and ankle during the first 6 minutes of treadmill running, after which no significant differences were found. The results from these studies suggest that a treadmill familiarization period of 6 minutes is required in order to obtain similar measurements from treadmill and overground locomotion trials.

Many studies have found evidence that suggests there are in fact differences between treadmill and overground locomotion. Differences have been found with respect to kinematics, electromyography (EMG), temporal variables and kinetics. Nelson et al. (Nelson, Dillman, Lagasse and Bickett, 1972) conducted one of the first studies which explored the biomechanics of experienced runners during treadmill and overground trials. They found kinematic differences between these two modes of running. Specifically, treadmill running was associated with longer periods of support, lower vertical velocity and less variable vertical and horizontal velocities. Nigg et al. (Nigg, De Boer and Fisher, 1995) investigated leg kinematics and found that individuals land with flatter feet during treadmill running when compared to overground running. They also found inconsistent patterns in lower extremity kinematics depending on landing style, running speed, and shoe and treadmill situations. In 1998, Wank et al. (Wank, Frick and Schmidtbleicher, 1998) found differences in the kinematics and muscle activities between overground and treadmill running. In this study treadmill trials were found to increase step frequency and to decrease step length, shoe sole angle at impact and knee joint angle at impact and stance phase. Differences with EMG were also seen among the treadmill trials. Specifically, a decrease in vastus lateralis activity during ground contact and an increase in biceps femoris activity during stance phase and the first part of the swing phase as well as an increase in rectus femoris activity during hip flexion. Alton et al. (Alton, Baldey, Caplan and Morrissey, 1998) found differences in leg joint kinematics and temporal variables with treadmill and overground walking. Significant increases were seen during treadmill walking in cadence, hip range of motion and maximum hip flexion joint angle whereas a significant decrease was seen in stance time. Kinetic differences have been found between treadmill and overground locomotion. White et al. (White, Yack, Tucker and Lin, 1998) looked at vertical ground reaction forces during treadmill and overground walking at slow, normal and fast speeds. They found significant force magnitude differences during normal and fast walking trials with an increase in vertical force (5-9%) during midstance and a decrease in peak vertical force during late stance with treadmill walking. Milgrom et al. (Milgrom, Finestone, Segev, Olin, Arndt and Ekenman, 2003) designed a study to determine if the kinetics of treadmill and overground running are reflected in differences in tibial strains and strain rates. This study found that axial compression and tension strains and strains rates were 48-285% higher among overground runners and suggested that treadmill runners were at a lower risk of developing tibial stress fractures.

It is apparent that the current literature regarding the accuracy of treadmills in measuring overground locomotion remains quite controversial. Many studies have shown that treadmills do not alter typical overground gait mechanics when examining angular kinematics and temporal and mechanical variables. Conversely, other studies suggest that treadmills do in fact alter typical overground locomotion particularly with respect to vertical force, electromyography, range of motion, lower extremity kinematics, time to treadmill familiarization and oxygen debt. To our understanding, there has been little research done to test the accuracy of treadmills in representing overground running with respect to maximum rearfoot motion and maximum internal tibial rotation.

Many factors may explain the controversial results seen in the literature. First, the above studies did not control for foot type. As a result the recruited sample may have consisted of participants with neutral, flat and high arched feet. The kinematics associated with each foot type may be remarkably different and therefore it is understandable why the overall kinematic results of treadmill and overground locomotion remain inconclusive. Second, footwear was worn by the participants in all of the previous studies with one exception. In some studies the participants wore their own shoes while in other studies the shoes were supplied, ensuring that all footwear worn was similar. However, if all foot types were not similar, it may have been the shoes that were altering typical gait and not the treadmill. Even if proper footwear was matched to the participant's foot type studies have shown that footwear may change the normal movement of the foot within the shoe during gait. Having the participants perform the trials barefoot may control for the effects of footwear on normal gait mechanics. Therefore further research is required, that controls for foot type and footwear, in order to confirm the use of treadmills in accurately representing overground running, particularly with respect to maximum rearfoot motion and maximum internal tibial rotation variables.

1.6 Research goals and hypotheses

The main research goals examined in this thesis:

1. To determine if kinematic differences exist between treadmill and overground running with respect to maximum rearfoot motion and maximum internal tibial rotation.
2. To determine if individuals with FFF demonstrate an increase in maximum rearfoot motion and maximum internal tibial rotation during treadmill running when compared to individuals with a subtalar neutral foot type.
3. To determine the mechanical effects of support under the medial longitudinal arch of the foot on maximum rearfoot motion and maximum internal tibial rotation during treadmill running among individuals with FFF.

The main hypotheses of this thesis:

1. No significant differences exist between treadmill and overground running with respect to maximum rearfoot motion and maximum internal tibial rotation.
2. Individuals with FFF demonstrate an increase in maximum rearfoot motion and maximum internal tibial rotation during treadmill running when compared to individuals with subtalar neutral foot type.

3. Arch supports under the medial longitudinal arch of the foot significantly reduce maximum rearfoot motion and maximum internal tibial rotation during treadmill running among individuals with FFF.

Chapters 2 and 3 will further investigate these research goals in an attempt to better understand 1) the applicability of the results from treadmill studies and 2) the running mechanics associated with FFF and the mechanism by which foot orthotics provide symptomatic relief. Chapter 4 will discuss the results of these studies and their clinical implications.

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CHAPTER 2: KINEMATIC DIFFERENCES BETWEEN TREADMILL AND OVERGROUND RUNNING

2.1 Abstract

The treadmill is a common instrument used to measure the lower extremity kinematics during running as repetitive gait cycles can be collected while controlling for velocity and incline. Previous studies involving gait research have been conducted with participants running on a treadmill or across an overground runway. In order to compare the results across previous studies, the assumption is made that similar gait mechanics occur during both treadmill and overground running. However this assumption remains controversial within the literature (Nelson, Dillman, Lagasse and Bickett, 1972; Ingen Schenau, 1980; Nigg, De Boer and Fisher, 1995; Wank, Frick and Schmidtbleicher, 1998; Schache, Blanch, Rath, Wrigley, Starr and Bennell, 2001; Lavcanska, Taylor and Schache, 2005). The purpose of this study was to compare the lower extremity kinematics between treadmill and overground running while controlling for foot type and footwear. A total of 19 healthy subjects with a subtalar neutral foot type performed a total of 12 barefoot running trials. Randomized running trials were completed on 2 different running surfaces (overground and treadmill) at 2 different speeds (2.0m/s and 3.0m/s). Measures of maximum rearfoot motion and maximum internal tibial rotation were calculated to determine if treadmill running mimics overground running. Running surface did not significantly affect rate of rearfoot angle ($p=0.1133$), maximum internal tibial rotation angle ($p=0.0517$) or rate of internal tibial rotation ($p=0.0549$). However, maximum rearfoot angle was significantly increased during treadmill running (9.7 ± 3.3deg) when compared to overground running (8.8 ± 2.9deg) ($p=0.045$) although this

difference was less than 1 degree. The results indicate that treadmills can confidently be used to measure lower extremity kinematics during overground running. However, careful interpretation should be employed when examining the magnitude of the maximum rearfoot angle obtained within treadmill studies.

2.2 Introduction

Interest in the underlying mechanics of running was ignited by the exponential growth of running as a recreational activity during the 1960's and 1970's (McClay, 2000). As the number of running related injuries increased, researchers began to investigate the mechanics of running in order to better understand the mechanisms associated with injury. As research studies emerged it became apparent that different methodologies were being utilized. More specifically, in some studies running trials were completed along an overground runway whereas in other studies the running trials were completed on a treadmill. Due to the many advantages associated with using a treadmill it is not surprising that many studies chose this method. Within a laboratory, treadmills require less space and allow for the collection of more gait cycles when compared to overground runways. In addition, treadmills allow for better control over incline, velocity and acceleration when compared to overground runways. However, in order to make comparisons between treadmill and overground studies, the assumption is made that treadmills do not alter typical running mechanics. Additionally, this assumption is of clinical importance since many health professionals use treadmills when diagnosing abnormal running mechanics.

Several studies have been conducted in order to test the accuracy of treadmills in representing overground running. The results have been controversial with some studies supporting and other studies challenging the above assumption. No significant differences have been found between treadmills and overground runways with respect to pelvis, hip, knee and ankle kinematics during running (Lavcanska, Taylor and Schache, 2005) or knee kinematics and temporal-distance gait measurements during walking (Matsas, Taylor and McBurney, 2000) assuming adequate time for treadmill familiarization, and three-dimensional kinematic patterns of the lumbo-pelvic-hip complex during running (Schache, Blanch, Rath, Wrigley, Starr and Bennell, 2001). Further, the use of a moving coordinate system to analyze data has been shown to result in no mechanical differences between overground and treadmill running (Ingen Schenau, 1980) assuming the treadmill belt speed is held constant. In contrast, other studies have shown kinematic differences between treadmill and overground runways. Treadmill running has been associated with longer periods of support and lower vertical velocity (Nelson, Dillman, Lagasse and Bickett, 1972), an increase in step frequency and a decrease in step length, shoe sole angle and knee joint angle at impact (Wank, Frick and Schmidtbleicher, 1998) and a flatter foot during stance (Nigg, De Boer and Fisher, 1995) when compared to overground running.

Many factors may explain the controversial results seen within the literature including footwear and foot type. Of the reviewed articles, all but one study involved running trials while wearing footwear. Variations within types of footwear included the standard lab shoe (Milgrom, Finestone, Segev, Olin, Arndt and Ekenman, 2003), the participants own personal shoe (Frishberg, 1983; Wank, Frick and Schmidtbleicher, 1998; Schache, Blanch, Rath, Wrigley, Starr and Bennell 2001), both standard and personal shoes (Nigg, De Boer and Fisher, 1995) or the type of footwear was not specified (Nelson, Dillman, Lagasse and Bickett, 1972; Matsas, Taylor and McBurney, 2000; Lavcanska, Taylor and Schache, 2005). Since the use of footwear has been shown to inaccurately represent the kinematic measurement of the foot (Nigg, 1986; Stacoff, Reinschmidt and Stussi, 1992) and the normal movement of the foot itself (Stacoff, Kaelin, Stuessi and Segesser, 1989; Stacoff, Kalin and Stussi, 1991), the kinematic results obtained from these studies may not be accurate. In addition, the previously mentioned studies did not control for foot type and thus, study populations may have included individuals with neutral, high and low arched feet. The kinematics associated with each foot type may have been remarkably different, and when compiled together, may explain the inconclusive results seen within the literature. According to our understanding, there has been no research to date that compares rearfoot and internal tibial rotation angles and rates between treadmill and overground running while controlling for footwear and foot type.

2.2.1 Specific aims

The purpose of this study was to investigate the lower extremity kinematics associated with treadmill and overground barefoot running among individuals with a subtalar neutral foot type. Researchers and clinicians typically examine the movement of the rearfoot to predict the movement of the underlying subtalar joint in order to diagnose abnormal running mechanics. If similar rearfoot and tibial rotation angles and rates are observed between these two running surfaces, then it can be postulated that treadmills accurately represent overground running. However, if lower extremity kinematic differences exist between these two running surfaces, then clinicians and researchers may need to cautiously interpret their results when attempting to predict overground running mechanics from treadmill running observations.

2.2.2 Hypotheses

H_O: No significant differences exist between treadmill and overground barefoot running with respect to rearfoot and internal tibial rotation angles and rates among individuals with a subtalar neutral foot type.

H_A: Treadmill barefoot running will differ significantly from overground barefoot running with respect to rearfoot and internal tibial rotation angles and rates among individuals with a subtalar neutral foot type.

2.3 Method

2.3.1 Participants

A total of 19 healthy individuals (mean ± standard deviation: age, 21.8 ± 3.2 years; height, 172.1 ± 9.8 cm; weight, 70.5 ± 13.7 kg; 10 women, 9 men) participated in this study. This study was reviewed and approved by the University Research Ethics Board at Wilfrid Laurier University. Study participants, recruited from the university population, were informed of the requirements and signed consent forms prior to testing. Individuals were deemed eligible to participate in this study if they met our predetermined subtalar neutral foot criteria.

Subtalar neutral foot type was defined as normal active range of motion (ROM) of the lower extremity, typical pelvic and knee alignment during static and dynamic visual inspections, a rearfoot angle between 4 and 6 deg and the presence of a medial longitudinal arch while sitting and standing (Appendix 2.1). Rearfoot angle was determined using a goniometer to measure the angle that resulted from the intersection of two lines: one line connected the mid calf and the Achilles tendon and the second line connected the superior and inferior aspects of the calcaneus (Figure 2.1). All participants had a rearfoot angle while standing that correlated with a neutral foot type (range 3-6 deg), with the group mean ± standard deviation being 4.5 ± 0.95 deg. In addition, participants completed a screening questionnaire (Appendix 2.2) and were excluded from this study if they had any neurological or physical condition that affected the use of their lower extremities.

Based on our criteria, all subjects demonstrated a subtalar neutral foot type. Only one subject demonstrated genu varum (bow-legged spanning 5 finger width) which was classified as atypical knee alignment. Running observations indicated that 16 participants initially struck the ground with their heel while the remaining subjects were mid/forefoot strikers.

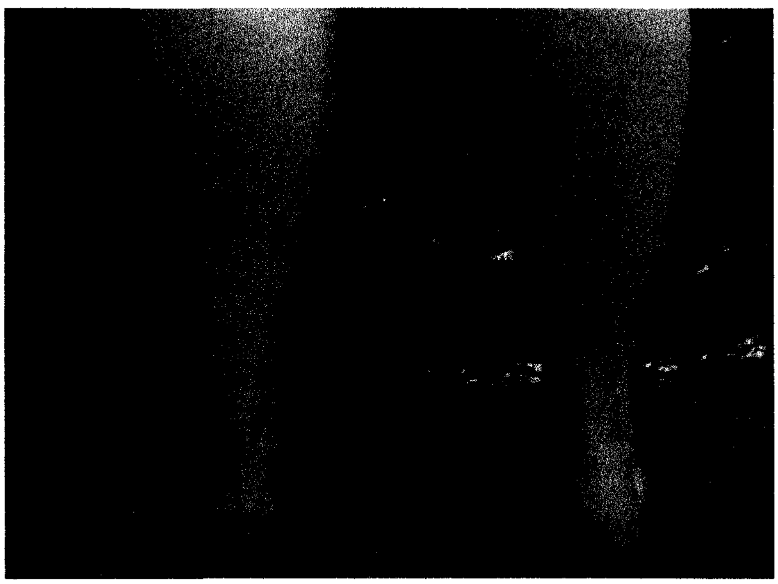

Figure 2.1: Rearfoot angle as determined by the difference between the angle of the leg and the angle of the calcaneus (RF angle = $\theta_{leg} - \theta_{calcaneus}$).

2.3.2 Experiment set-up and data collection

The laboratory used for data collection measured 10m X 8m and was set up for overground and treadmill running trials as illustrated in Figures 2.2 and 2.3, respectively. The overground runway was approximately 14m in length and was positioned diagonally across the laboratory floor to allow for maximal running distance while ensuring foot strike with the force plate.

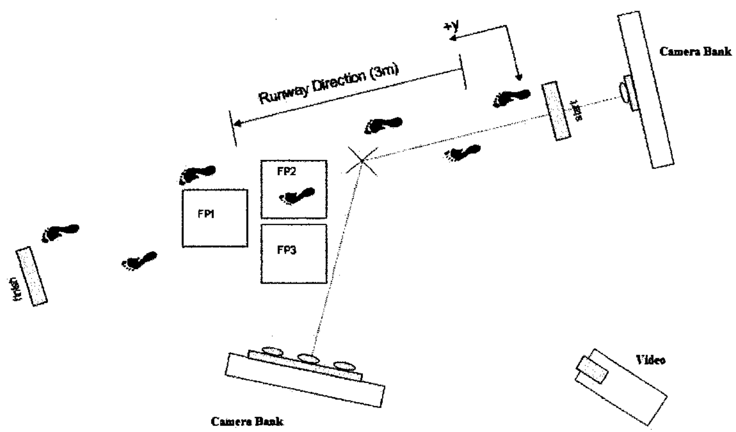

Figure 2.2: Laboratory set-up for overground running trials.

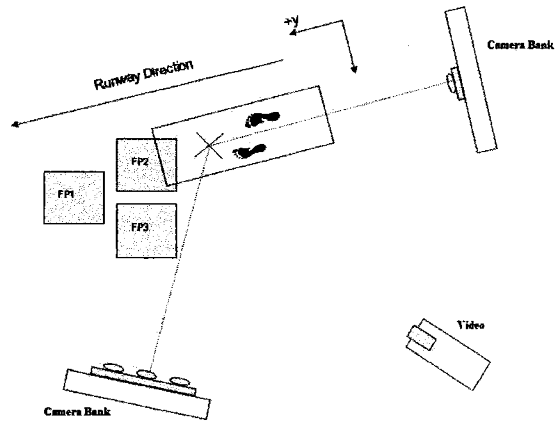

Figure 2.3: Laboratory set-up for treadmill running trials.

Kinematic data was collected at 100Hz using 2 OptoTrak 3020 camera banks (Northern Digital Inc., Waterloo, Canada) and the ToolBench computer software. A total of 20 infrared light-emitting diodes (IRED's) were placed on each rearfoot (4) and tibia (6) in order to track the motion of these segments (Figure 2.4).

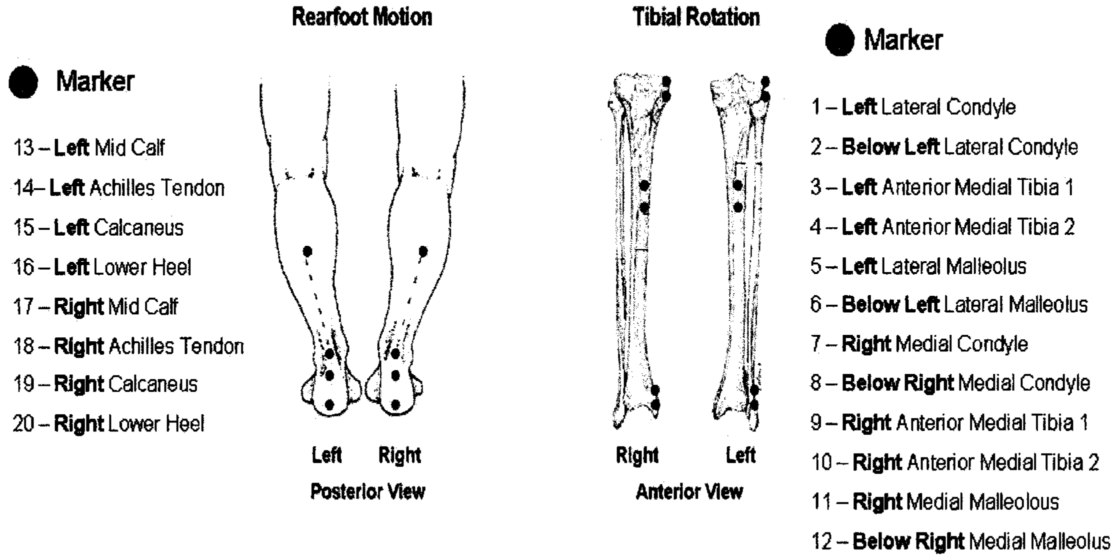

Figure 2.4: IRED placement on the rearfoot and tibia in order to track rearfoot angle and internal tibial rotation during running.

The treadmill (Figure 2.5) used to complete the treadmill running trials was a Precor M9.21si (Precor Inc., Bothell, WA USA). The dimensions of the running surface was 129.5cm X 43cm with the overall length, width (including handrails), and height of the treadmill measuring 170cm, 71cm and 111.8cm respectively. It contained a 2.0 hp motor.

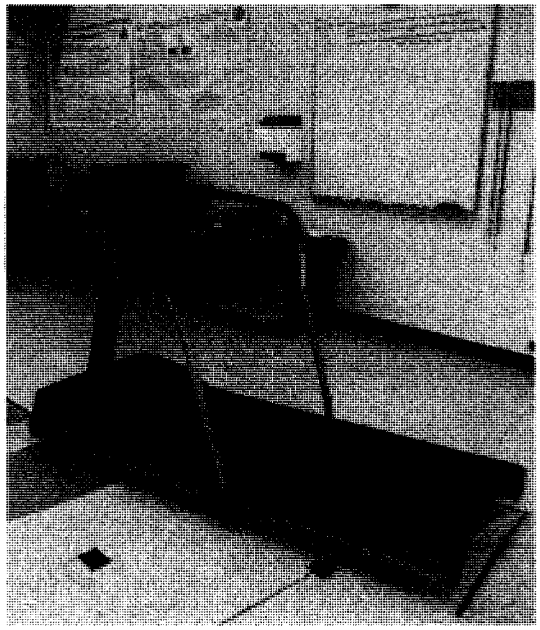

Figure 2.5: Treadmill used in this study (M9.21si, Precor Inc., Bothell, WA USA).

2.3.3 Procedure

This study consisted of two sessions. During Session 1 the subtalar neutral foot criteria and the exclusion questionnaire were completed. If the participants were deemed eligible based on the criteria and the questionnaire they were invited back to complete Session 2.

During Session 2 the kinematic markers were applied to the rearfoot and tibia of each leg on the participant as illustrated in Figure 2.4. They were then given as many overground and treadmill practice trials as required in order to feel comfortable running at 2.0m/s and 3.0m/s. A spotter stood beside them throughout all running trials to ensure their safety. Once participants felt comfortable to begin they completed the barefoot running trials under 4 conditions: overground 2.0m/s; overground 3.0m/s; treadmill 2.0m/s; treadmill 3.0m/s. A total of 10 participants completed the overground 2.0m/s trial first and continued through the conditions as described above. The remaining 9 participants started with the treadmill 3.0m/s trial and moved in reverse order thus finishing with the overground 2.0m/s trial. To ensure accurate running velocity during the overground trials, a 3m distance was marked out on the floor. Using a stopwatch, a lab assistant recorded the amount of time it took the participant to move across the 3m distance. For example, a time of 1.5s indicated a speed of 2.0m/s whereas a time of 1.0s indicated a speed of 3.0m/s. In addition, successful completion of the overground trials included striking the force plate with the left foot while running. This allowed for consistent determination of heel contact among all subjects. A total of 3 successful overground running trials at each speed were completed for each participant.

Before beginning the treadmill running trials, the treadmill was positioned with the front, left corner resting on the force plate. Again, the vertical force information allowed for the determination of heel contact with the force from the left foot having a higher magnitude than the right foot. Once the participant felt ready to begin, the treadmill speed was brought up to either 2.0m/s or 3.0m/s. A total of three 5 second trials were collected before the treadmill speed was returned to a comfortable walking pace as set by the participant. On the participants command, the treadmill speed was again increased to record three consecutive trials at the remaining speed. Once these trials were collected the speed was decreased to a comfortable walking pace until the participant indicated that they had completed sufficient cool down time and were ready to step off. Following completion of all barefoot running trials, the treadmill was removed from the runway and the participant stood quietly on the force plate in the direction of the runway while a 5 second static trial was recorded. Thus, Session 2 involved the completion of 13 trials (12 running and 1 static).

2.3.4 Data analysis and statistics

The four conditions in this protocol allowed for the analysis of whether lower extremity kinematics differed between overground and treadmill running at 2 different speeds (Figure 2.6). The independent variables in this study were surface (overground or treadmill), velocity (2.0m/s or 3.0m/s) and foot (right or left). The dependent kinematic variables analyzed in this study were rearfoot motion (used to measure foot pronation) and internal tibial rotation. Please refer to Table 2.1 for a description of each kinematic variable analyzed in this study. A priori analysis of estimation of sample size was conducted prior to subject recruitment in order to ensure that a statistical power of 0.8 was achieved.

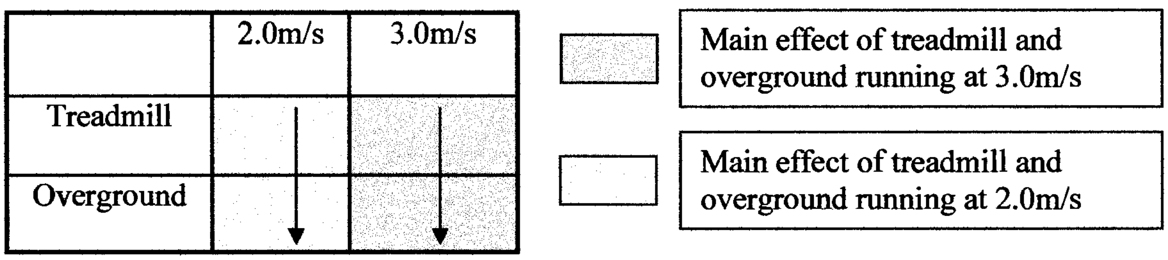

Figure 2.6: Repeated measures analysis of variance between treadmill and overground running.

Rearfoot angle is commonly used to estimate foot pronation due to the difficulty in measuring this variable. In this study, rearfoot angle was calculated using the four rearfoot kinematic markers (Figure 2.4). The top two markers allowed for the determination of the absolute leg angle relative to the horizontal; the bottom two markers were used to determine the absolute calcaneal angle relative to the horizontal. The relative angle of the rearfoot was determined by subtracting the absolute angle of the leg from the absolute angle of the calcaneus:

$$\text{Rearfoot angle} = \theta_{leg} - \theta_{calcaneus}$$

The second dependent variable, internal tibial rotation, was tracked by six kinematic markers that were placed on the tibia (Figure 2.4). Computer software (Visual 3D) was used to create a model of the right and left tibia for each subject. This model was created from a standing trial and then assigned to the running trials. Thus, this study analyzed internal tibial rotation relative to the static position of the tibia. Internal tibial rotation was calculated within the Visual 3D software and was defined as rotation around the vertical (z) axis in the transverse plane. Both rearfoot angle and internal tibial rotation were calculated for every frame within each 5 second trial.

Table 2.1: A description of the kinematic variables analyzed in this study. All angles and rates were calculated for every frame during each 5 second trial.

Kinematic Variable	Description
Maximum Rearfoot Angle	Rearfoot angle (RF_θ) was determined by calculating the difference between the leg and calcaneal angles. Maximum rearfoot angle (Max RF_θ) was defined as the maximum rearfoot angle achieved during the stance phase minus the rearfoot angle at heel contact: $\text{Max } RF_\theta = RF_{\theta \text{ max stance}} - RF_{\theta \text{ heel contact}}$
Rate of Rearfoot Angle	Rate of rearfoot angle (RF_{rate}) was defined as the rate at which the rearfoot achieved the maximum rearfoot angle during stance. It was calculated by dividing the Max RF_θ by the difference in time from heel contact to maximal stance: $RF_{rate} = \dfrac{\text{Max } RF_\theta}{t_{RF\theta \text{ max stance}} - t_{RF\theta \text{ heel contact}}}$

Maximum Internal Tibial Rotation Angle	Internal tibial rotation angle (ITR_θ) was defined as rotation around the vertical (z) axis in the transverse plane and was calculated using Visual 3D software. Maximum ITR_θ was defined as the difference between ITR_θ at heel contact and maximal stance: $\text{Max } ITR_\theta = ITR_{\theta\ \text{max stance}} - ITR_{\theta\ \text{heel contact}}$
Rate of Internal Tibial Rotation	Rate of internal tibial rotation (ITR_{rate}) was defined as the rate at which the tibia achieved the maximum internal rotation angle during stance. It was calculated by dividing the Max ITR_θ by the difference in time from heel contact to maximal stance : $ITR_{rate} = \dfrac{\text{Max } ITR_\theta}{t_{ITR\theta\ \text{max stance}} - t_{ITR\theta\ \text{heel contact}}}$.

A program was written in Visual Basic specifically for the analyses in this study. This program allowed for the selection of OptoTrak, force plate and Visual 3D data and then displayed them in a graph. This graph showed the position of the right (marker 20) or left (marker 16) foot as well as the vertical force. The vertical force data was collected during this study in order to more accurately determine when heel contact and toe off had occurred. After selecting an area of the graph that correlated with an increase in vertical force, rearfoot and internal tibial rotation curves were produced. From these curves, maximum angles and rates of both the rearfoot and tibia were determined. During each trial, one stance phase was selected and analyzed for both the right and left foot.

The results were analyzed using the SAS computerized statistical package. A three factor (2 conditions X 2 speeds X 2 feet) within-subject repeated measures ANOVA was used to analyze the results with the a priori alpha set at 0.05. All rearfoot and tibial rotation outliers with a standard residual greater than 2.0 were investigated. A total of 11 outliers had missing data during stance ranging from 10 frames to the entire stance phase and therefore were excluded from the analysis.

2.4 Results

The purpose of this study was to determine if treadmills accurately represent overground running with respect to rearfoot motion and internal tibial rotation. This section describes the results found for each dependent variable: 1) maximum rearfoot angle, 2) rate of rearfoot angle, 3) maximum internal tibial rotation angle, and 4) rate of internal tibial rotation. The result of each variable begins with a description of how running surface, running speed and foot affected that particular variable as well as any interactions that occurred. The figures illustrate the effects of running surface and running speed on the dependent variable of interest. Following these figures are graphs which illustrate the interactions for each variable if they were found to be significant. The tables at the end of this section summarize the effects of running surface (Table 2.2), running speed (Table 2.3) and foot (Table 2.4) on each of the dependent variables.

2.4.1 Maximum rearfoot angle

The type of running surface had a significant effect on maximum rearfoot angle ($p=0.045$). More specifically, the maximum rearfoot angle was significantly less while running overground (8.8 ± 2.9 deg) when compared to running on a treadmill (9.7 ± 3.3 deg) (Table 2.2). Running speed also had a significant effect on maximum rearfoot angle ($p=0.048$) as running at 2.0m/s and 3.0m/s was associated with maximum rearfoot angles of 8.8 ± 3.0 deg and 9.7 ± 3.2 deg, respectively (Table 2.3). A significant difference with respect to maximum rearfoot angle was also observed between feet ($p<0.0001$). For example, the left foot had a smaller maximum rearfoot angle when compared to the right foot (8.1 ± 2.3 deg versus 10.4 ± 3.4 deg, respectively) (Table 2.4). Figure 2.7 illustrates the effects of running surface and running speed on maximum rearfoot angle.

No significant interactions occurred with the maximum rearfoot angle variable.

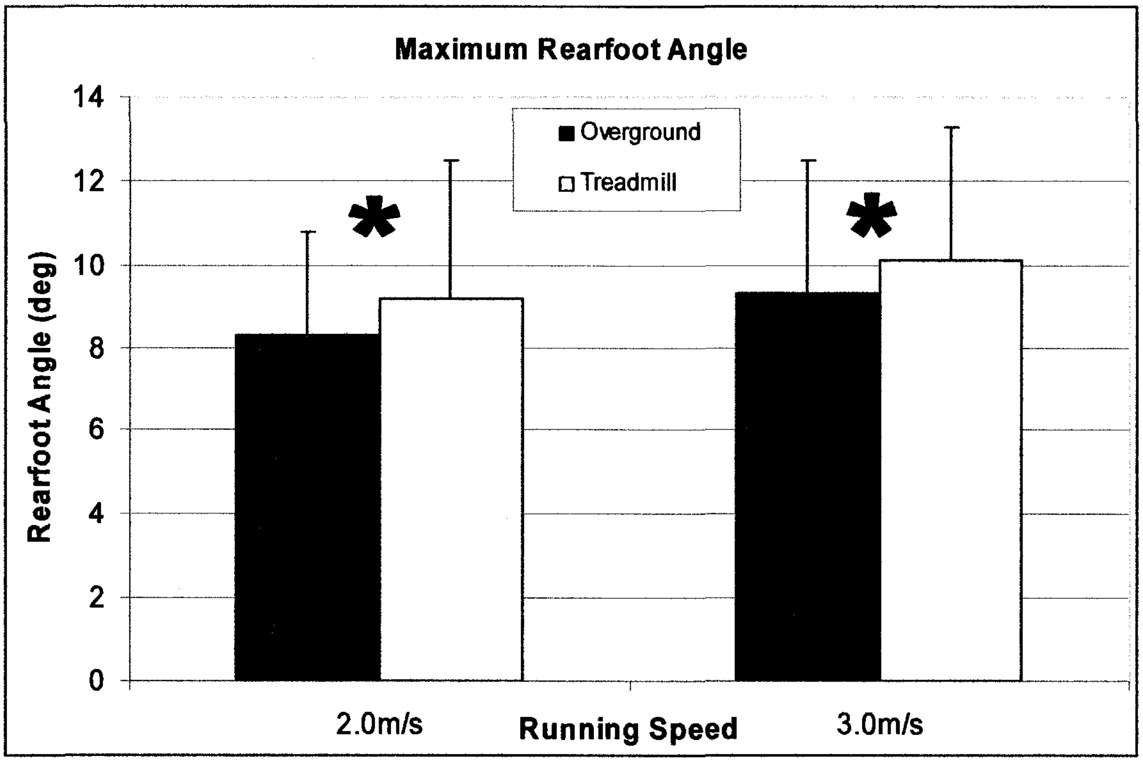

Figure 2.7: Maximum rearfoot angle (mean + standard deviation) obtained while running overground and on a treadmill at 2 different speeds. The symbol '*' denotes significance.

2.4.2 Rate of rearfoot angle

The type of running surface did not have a significant effect on the rate of rearfoot angle ($p=0.1133$) as overground and treadmill running demonstrated values of 92.2 ± 49.9 deg/sec and 100.9 ± 46.8 deg/sec, respectively (Table 2.2). Rate of rearfoot angle was significantly affected by running speed ($p=0.0001$). Specifically, running at 2.0m/s and 3.0m/s was associated with values of 83.9 ± 43.6 deg/sec and 109.7 ± 49.8 deg/sec, respectively (Table 2.3). No significant differences in the rate of rearfoot angle were present between left (101.8 ± 54.3 deg/sec) and right (91.3 ± 40.8 deg/sec) feet ($p=0.2087$) (Table 2.4). Figure 2.8 illustrates the effects of running surface and running speed on the rate of rearfoot angle.

There was a significant interaction between running speed and foot ($p=0.0037$). More specifically, the rate of rearfoot angle was significantly less within the right foot (96.4 ± 39.4 deg/sec) when compared to the left foot (123.1 ± 55.4 deg/sec) while running at 3.0m/s only (Figure 2.9).

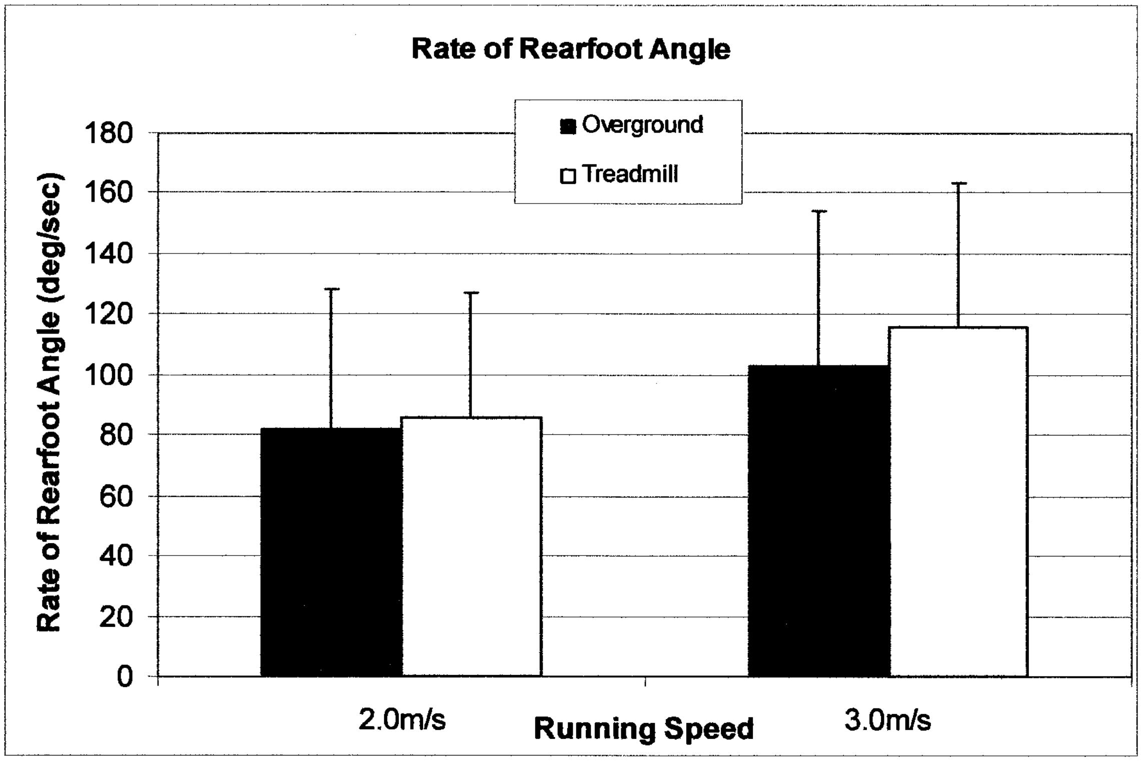

Figure 2.8: Rate of rearfoot angle (mean + standard deviation) obtained while running overground and on a treadmill at 2 different speeds.

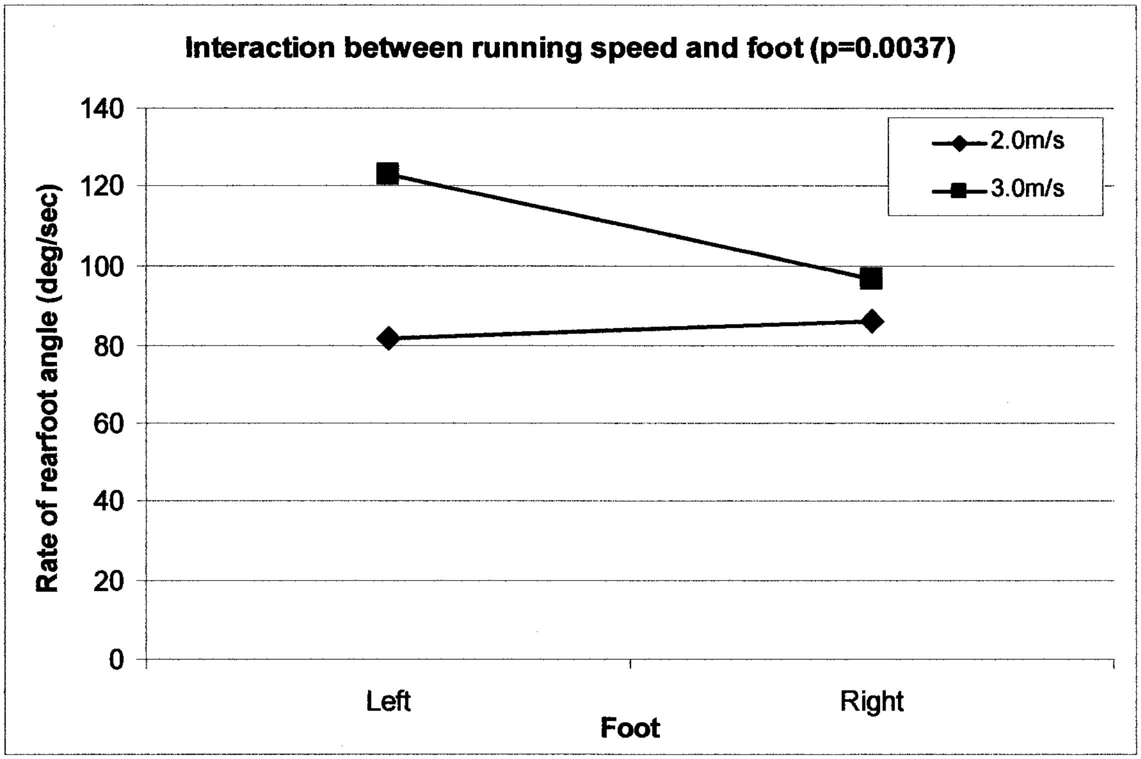

Figure 2.9: The interaction between running speed and foot (p=0.0037) on the rate of rearfoot angle. Mean values presented; standard deviations are presented on page 65.

2.4.3 Maximum internal tibial rotation angle

Maximum internal tibial rotation angle was not significantly affected by running surface ($p=0.0517$) as overground and treadmill running produced maximum internal tibial rotation angles of 12.2 ± 10.1 deg and 13.5 ± 8.4 deg, respectively (Table 2.2). Running speed did not significantly affect maximum internal tibial rotation angles ($p=0.1174$) as 2.0m/s and 3.0m/s demonstrated values of 13.6 ± 9.7 deg and 12.2 ± 8.6 deg, respectively (Table 2.3). Maximum internal tibial rotation angle was not significantly affected by foot ($p=0.7314$) as the left and right foot produced maximum internal tibial rotation angles of 13.6 ± 12 deg and 12.2 ± 4.5 deg, respectively (Table 2.4). Figure 2.10 illustrates the effects of running surface and running speed on maximum internal tibial rotation angle.

There was a significant interaction between running surface and foot ($p=0.0339$). More specifically, maximum internal tibial rotation angle was significantly less within the right foot (12.6 ± 4.1 deg) when compared to the left foot (14.4 ± 11.3 deg) during treadmill running only (Figure 2.11).

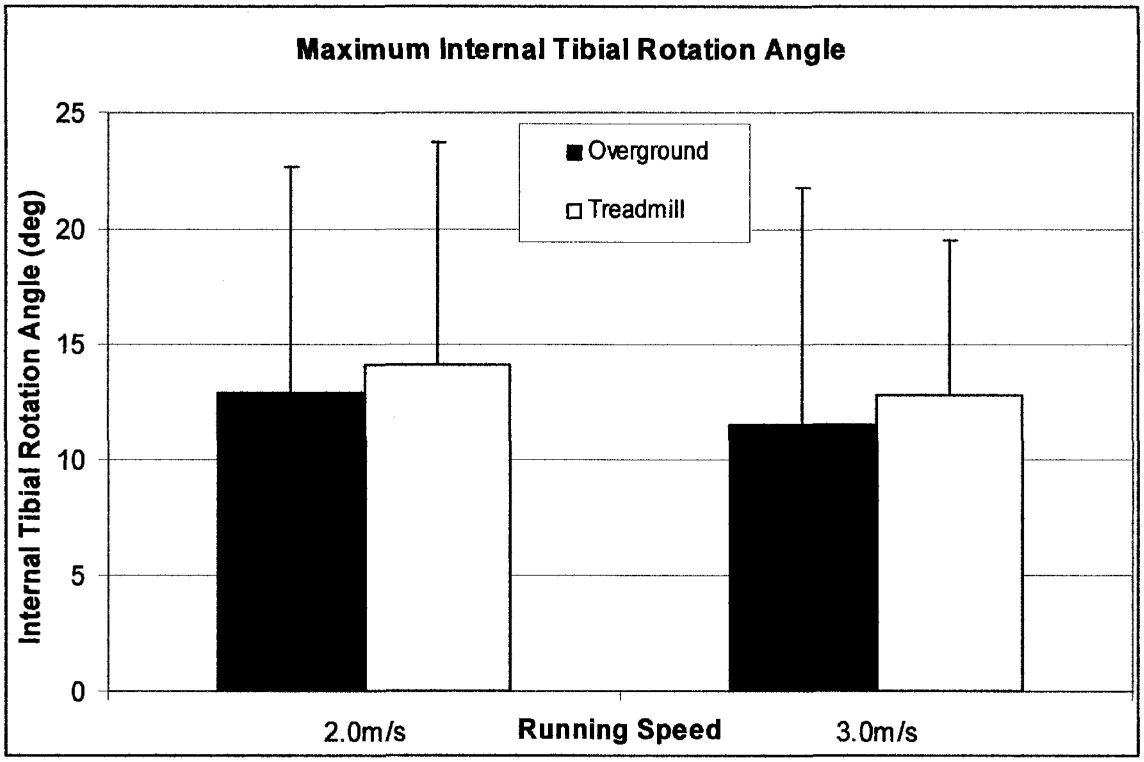

Figure 2.10: Maximum internal tibial rotation angle (mean + standard deviation) obtained while running overground and on a treadmill at 2 different speeds.

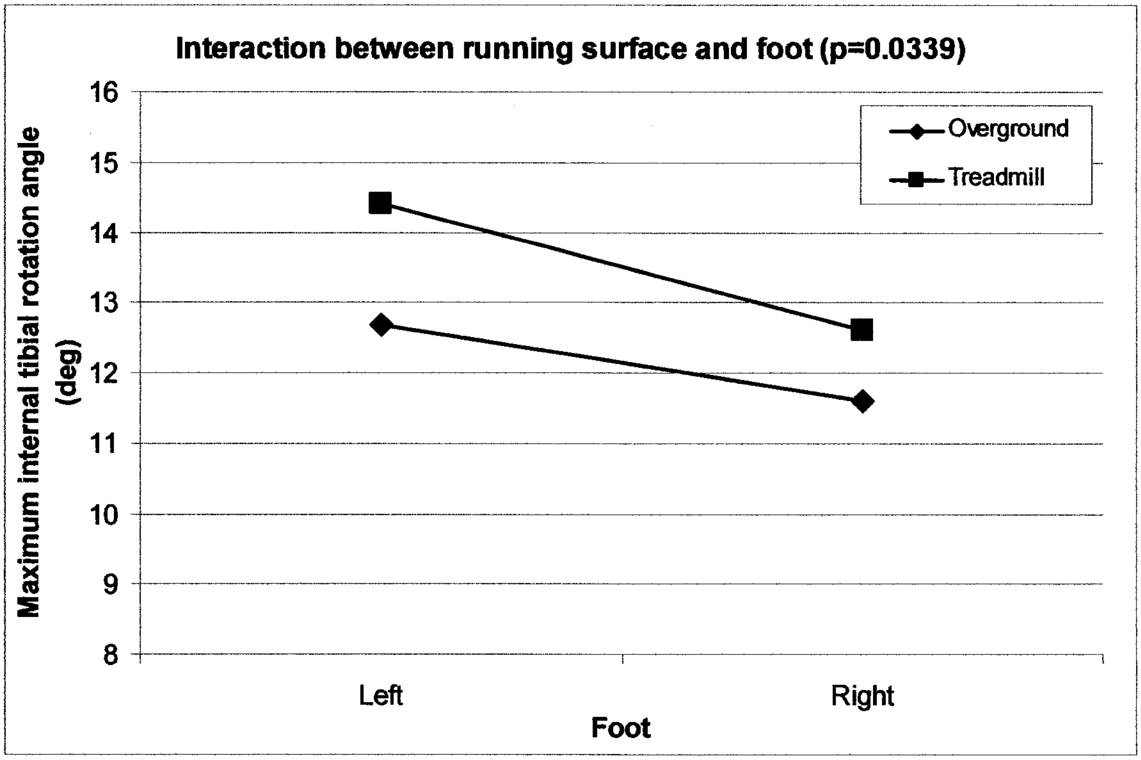

Figure 2.11: The interaction between running surface and foot (p=0.0339) on maximum internal tibial rotation angle (deg). Mean values presented; standard deviations are presented on page 68.

2.4.4 Rate of internal tibial rotation

Rate of internal tibial rotation was not significantly affected by running surface ($p=0.0549$) as overground and treadmill running produced values of 94.7 ± 77.3 deg/sec and 104.6 ± 69 deg/sec, respectively (Table 2.2). Running speed did not significantly affect the rate of internal tibial rotation ($p=0.5044$) as 2.0m/s and 3.0m/s demonstrated values of 96.5 ± 70.9 deg/sec and 104 ± 75.2 deg/sec, respectively (Table 2.3). Rate of internal tibial rotation was not significantly affected by foot ($p=0.9259$) as the left and right foot produced values of 102.0 ± 83.8 deg/sec and 98.0 ± 59.3 deg/sec, respectively (Table 2.4). Figure 2.12 illustrates the effects of running surface and running speed on the rate of internal tibial rotation.

There was a significant interaction between running speed and running surface ($p=0.0090$). More specifically, the rate of internal tibial rotation was significantly increased during treadmill running between speeds of 2.0m/s (97.4 ± 71 deg/sec) and 3.0m/s (112.5 ± 66.3 deg/sec) (Figure 2.13). A significant interaction was also demonstrated between running speed and foot ($p=0.0183$). More specifically, there was a significant decrease in the rate of internal tibial rotation with the right foot (93.7 ± 54.8 deg/sec) when compared to the left foot (99 ± 82.7 deg/sec) while running at 2.0m/s (Figure 2.14).

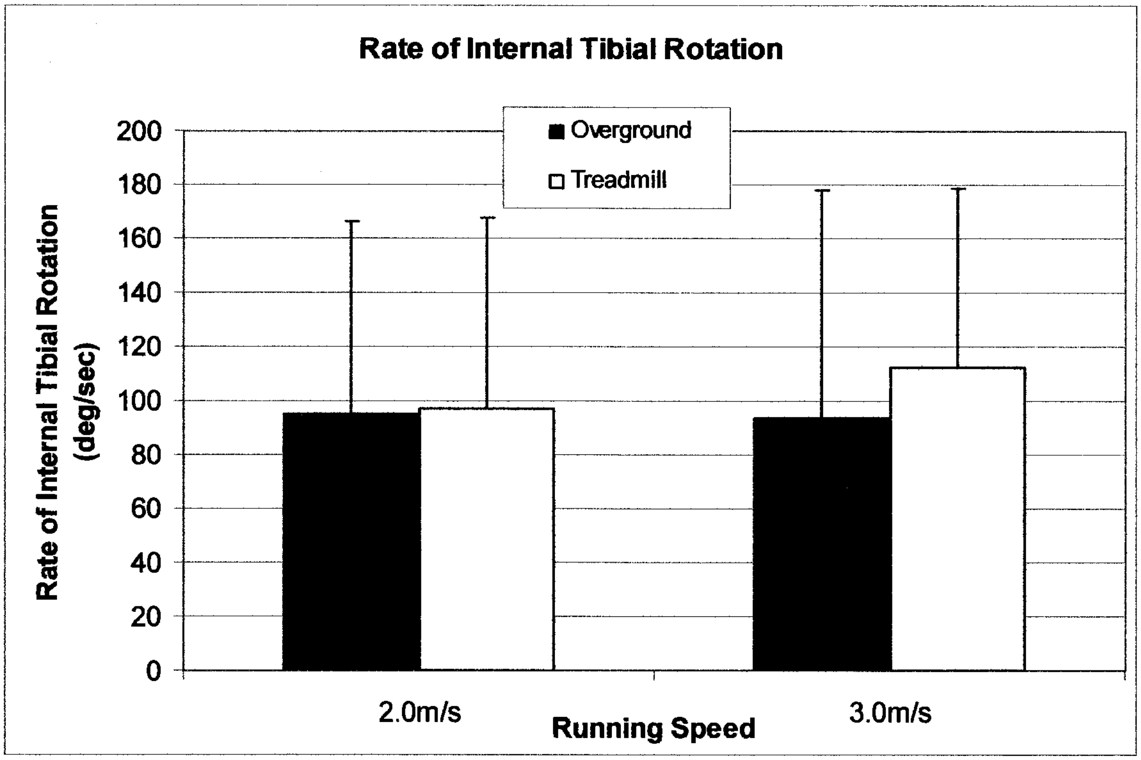

Figure 2.12: Rate of internal tibial rotation (mean + standard deviation) obtained while running overground and on a treadmill at 2 different speeds.

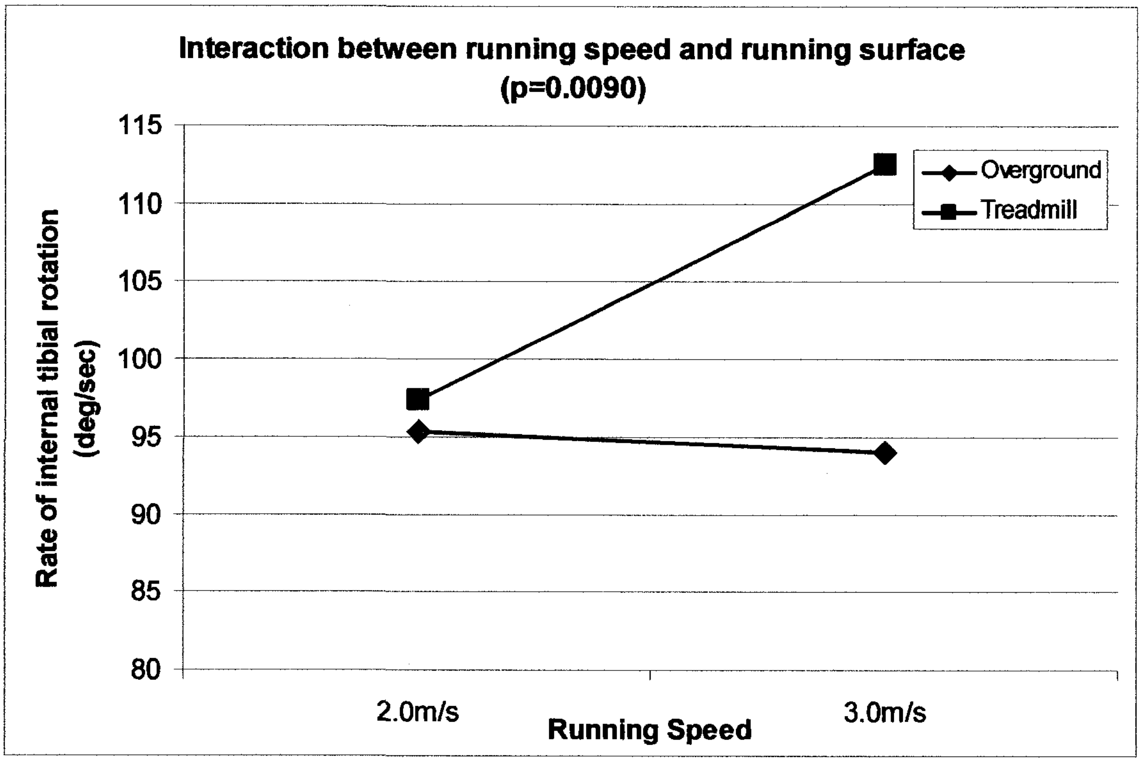

Figure 2.13: The interaction between running speed and running surface (p=0.0090) on the rate of internal tibial rotation (deg/sec). Mean values presented; standard deviations are presented on page 71.

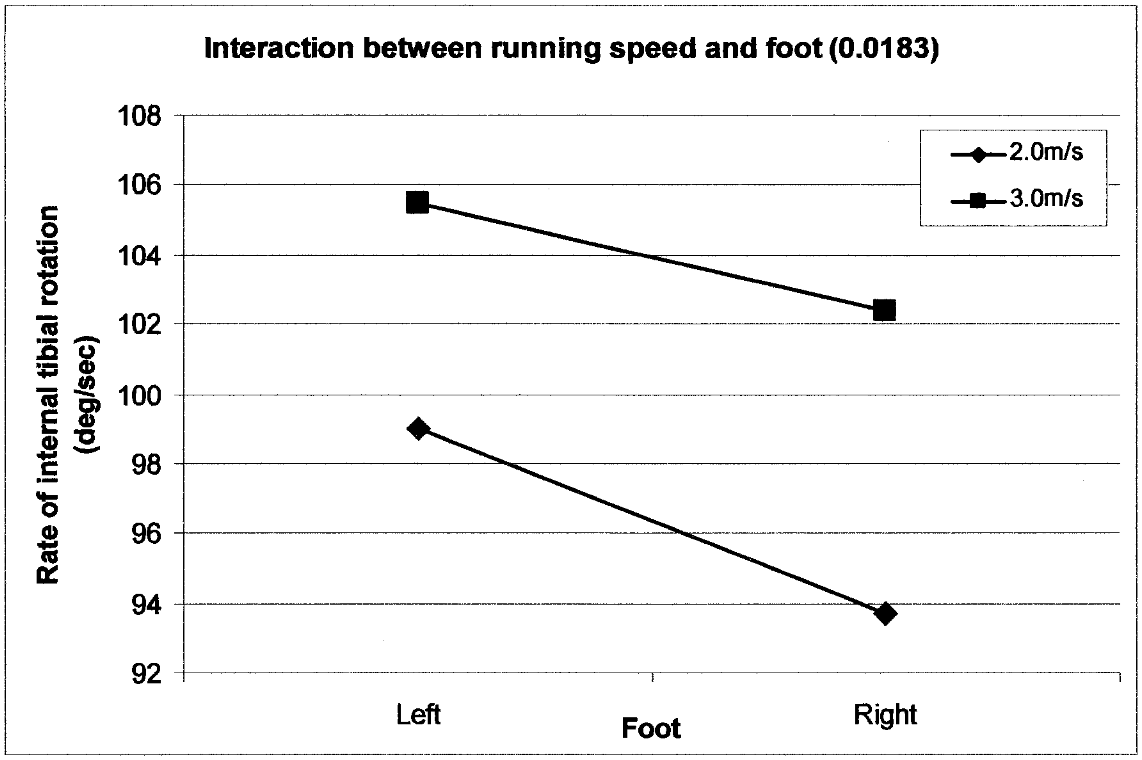

Figure 2.14: The interaction between running speed and foot ($p=0.0183$) on the rate of internal tibial rotation (deg/sec). Mean values presented; standard deviations are presented on page 71.

Table 2.2: The effects of surface type (overground and treadmill) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance ($p \leq 0.05$) between surface type.

	Overground	Treadmill
Max Rearfoot Angle (deg) *	8.8 (2.9)	9.7 (3.3)
Rate of Rearfoot Angle (deg/sec)	92.2 (49.9)	100.9 (46.8)
Max Internal Tibial Rotation Angle (deg)	12.2 (10.1)	13.5 (8.4)
Rate of Internal Tibial Rotation (deg/sec)	94.7 (77.3)	104.6 (69)

Table 2.3: The effects of running speed (2.0m/s and 3.0m/s) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance ($p \leq 0.05$) between running speed.

	2.0m/s	3.0m/s
Max Rearfoot Angle (deg) *	8.8 (3.0)	9.7 (3.2)
Rate of Rearfoot Angle (deg/sec) *	83.9 (43.6)	109.7 (49.8)
Max Internal Tibial Rotation Angle (deg)	13.6 (9.7)	12.2 (8.6)
Rate of Internal Tibial Rotation (deg/sec)	96.5 (70.9)	104 (75.2)

Table 2.4: The foot effect (left and right) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance (p≤0.05) between foot.

	Left Foot	Right Foot
Max Rearfoot Angle (deg) *	8.1 (2.3)	10.4 (3.4)
Rate of Rearfoot Angle (deg/sec)	101.8 (54.3)	91.3 (40.8)
Max Internal Tibial Rotation Angle (deg)	13.6 (12)	12.2 (4.5)
Rate of Internal Tibial Rotation (deg/sec)	102 (83.8)	98 (59.3)

2.5 Discussion

The objective of this research study was to investigate the accuracy of treadmills in representing overground running. We hypothesized that there would be no significant difference between treadmill and overground running with respect to lower extremity kinematic variables. The results of this study support this hypothesis in terms of the rate of rearfoot angle, maximum internal tibial rotation angle and the rate of internal tibial rotation. However, maximum rearfoot angle was significantly greater during treadmill running when compared to overground running, albeit less than 1 degree. This section begins with a discussion of the observed results with respect to rearfoot motion and internal tibial rotation. Then, the clinical applications and study limitations are presented followed by recommendations for future research and overall conclusions.

2.5.1 Rearfoot motion

The kinematic marker set-up and procedures used to measure rearfoot motion in this study has been used in previous research (Nike, 1989; Kernozek and Ricard, 1990; Perry and Lafortune, 1995; McClay and Manal, 1997; McClay and Manal, 1998; Hetsroni, Finestone, Milgrom, Ben-Sira, Nyska, Mann, Almosnino and Ayalon, 2008). The results of this study indicate that treadmills may slightly over predict maximum rearfoot angles when compared to overground running (overground, 8.8 ± 2.9 deg; treadmill, 9.7 ± 3.3 deg). Similar maximum rearfoot angles (range: 8.23-11.2 deg) have been reported in previous work involving treadmill and overground running (McClay and Manal, 1997; McClay and Manal, 1998; Nigg, Khan, Fisher and Stefanyshyn, 1998; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). From these studies, running trials performed on a treadmill elicited maximum rearfoot angles closer to the high end of the range (11.2 deg) whereas running trials performed overground elicited maximum rearfoot angles closer to the low end of the range (8.23 deg and 10.5 deg). In a particular study involving overground running trials at 2.5-3.0m/s, the maximum rearfoot angle of 8.23 deg was demonstrated which is very similar to the findings of the present study (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). Although significance was achieved with maximum rearfoot angle between overground and treadmill running, it is unknown whether this small difference (<1 deg) would be of clinical significance.

The current literature suggests that the type and size of the treadmill may have implications on the kinematic effects of locomotion. A study which compared different types of treadmills to overground sprinting kinematics indicated that torque treadmills may more accurately represent overground sprinting when compared to conventional motorized treadmills (McKenna and Riches, 2007). In addition, a narrow treadmill (34.5cm) with a hard surface has been shown to elicit more foot pronation during the initial portion of the stance phase while walking when compared to a wider treadmill (50.5cm) with a soft surface (Sajko and Pierrynowski, 2005). Although treadmill softness is not known, the width of the treadmill used in the current study was 43cm and thus, running surface width may partly be responsible for the increase in maximum rearfoot angle. Therefore, the significant result of maximum rearfoot angle may be associated with the size and type of the treadmill used in this study and may not be reflective of all treadmill running in general.

In terms of the rate of rearfoot angle, the results from this study indicate that treadmill running did not differ significantly from overground running (overground, 92.2

± 49.9 deg/sec; treadmill, 100.9 ± 46.8 deg/sec). This finding may be of greater importance in terms of injury prevention since increases in maximum rearfoot eversion velocity have been associated with overuse running injuries (Hreljac, Marshall and Hume, 2000). Previous work on rearfoot motion during running has reported values for the rates of rearfoot angle that are consistent with the values measured in this study. Specifically, while running at a similar speed (2.5-3.0m/sec) maximum rearfoot velocity ranged from 73.17 deg/sec to 157.41 deg/sec between subjects (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). At a slightly faster running speed (3.6m/sec) maximum rearfoot velocities ranging from approximately 55 deg/sec to 190 deg/sec were noted between subjects (MacLean, McClay Davis and Hamill, 2006).

2.5.2 Internal tibial rotation

The kinematic markers were placed at locations on the tibia that had minimal soft tissue between the bone and the skin and could be seen by the camera during the running trials. These concepts for optimal marker placement on the shank have been supported by previous work (Nigg, Khan, Fisher and Stefanyshyn, 1998; Bellchamber and van den Bogert, 2000). The method used to calculate internal tibial rotation in this study has been used in previous research (Perry, 1993). The results from this study indicate that treadmills accurately represent internal tibial rotation angle during overground running as no significant differences were found between these running surfaces (overground, 12.2 ± 10.1 deg; treadmill 13.5 ± 8.4 deg). Similar values have been reported within the literature. A study investigating the transverse rotation of the tibia during walking found values ranging from 3.5 - 14.4 deg (Lafortune, 1984). Nigg et al. (Nigg, Cole and Nachbauer, 1993) reported internal tibial rotation values of approximately 21 ± 8.4 deg during the stance phase of running. It has been speculated that approximately 6.9 deg of the total 21 deg may be a result of external tibial rotation during heel contact (Perry, 1993). Therefore, a more accurate value for their measured internal tibial rotation may be closer to 14 deg which is similar to the present study. Other research studies have measured internal tibial rotation relative to the foot (Nigg, Cole and Nachbauer, 1993; McClay and Manal, 1997; Nigg, Khan, Fisher and Stefanyshyn, 1998). The results from these studies demonstrate internal tibial rotation angles ranging between approximately 4.8 deg to 8.9 deg. These values, which are lower than those found in the present study, may be explained by their choice of reference frame. Since the foot and tibia move together during gait, internal tibial rotation relative to the foot may be less than if it was measured relative to a static structure. The present study measured internal tibial rotation relative to its static position which may explain the increased internal tibial rotation angles found in the present study.

The rate of internal tibial rotation did not differ significantly between treadmill and overground running (overground, 94.7 ± 77.3 deg/sec; treadmill, 104.6 ± 69.0 deg/sec). Thus, treadmills accurately represent the rate of internal tibial rotation during overground running. Although few studies have examined this variable, the findings from the present study appear to be lower than what has been previously shown. In a study involving overground running at 4.0m/s, a maximum tibial rotation velocity of approximately 185 deg/sec was demonstrated (Mundermann, Nigg, Humble and Stefanyshyn, 2003). This increase in velocity may be explained by the increase in running speed or due to the fact that the reported velocity may have included that of both internal and external tibial rotation.

This study demonstrated no significant differences between treadmill and overground running with respect to internal tibial rotation which is particularly interesting since running injuries have been shown to most commonly occur at the knee (Clement, Taunton, Smart and McNicol, 1981). Therefore, it can be postulated that treadmill running may not further increase the risk of developing running related knee injuries.

2.5.3 Clinical applications

The results from this study indicate that treadmills can be used to accurately represent the lower extremity kinematics associated with overground running. This instrument allows for the collection of repetitive gait cycles while controlling for velocity and incline when compared to overground runways and as a result may better contribute to our understanding of normal and abnormal running mechanics as well as optimal footwear design. It is currently unknown whether the small statistically significant increase in rearfoot angle during treadmill running would be of clinical significance. However, if clinical decisions are dependent on small changes in maximum rearfoot angle then cautious interpretation should be employed when using treadmills.

2.5.4. Limitations

Although the same number of stance phases were analyzed for both the overground and treadmill conditions there was a smaller number of acceptable foot contacts that occurred during overground running when compared to treadmill running. For example, as the participant ran across the overground runway only one acceptable gait cycle for both the right and left foot was recorded by the camera banks. Therefore, only one stance phase was analyzed for both the right and left foot during overground and treadmill running. This may have been a limitation since the lower extremity kinematics for that trial relied solely on one single stance phase as opposed to an average of multiple stance phases.

Another limitation may have occurred due to the size of the treadmill running surface used in this study. It has been postulated that a narrow (34.5cm) and soft treadmill running surface results in an increase in foot pronation during the initial portion of the stance phase when compared to a wide (50.5cm) and hard treadmill running surface (Sajko and Pierrynowski, 2005). Although the width of the current treadmill was 43cm, this may not have been wide enough to accommodate each participant's natural stride. As a result, a narrower running surface may have caused a disruption to the normal movement pattern during running resulting in an increase in maximum rearfoot angle.

Although this study attempted to recruit individuals who exhibited a heel to toe running pattern, it was determined during testing that 3 subjects initially struck the ground with their forefoot. This may have had implications on the analysis procedures since different running patterns are associated with forefoot strikers when compared to heel strikers (McClay and Manal, 1995). However, these individuals were consistent forefoot strikers during both overground and treadmill running trials therefore this may not have had an effect on the overall results.

2.5.5 Recommendations for future research

This was an initial study designed to test the accuracy of treadmills in representing overground running with respect to lower extremity kinematics. Suggestions for future research are listed below:

1. Further examine treadmill size, particularly running surface width, in comparison to overground running to determine an acceptable treadmill width threshold for accurate lower extremity measurement.
2. Evaluate the effects of unconventional treadmills (torque treadmills) on lower extremity kinematic variables during walking and running.
3. Include clinical populations in future research since their foot abnormalities are typically diagnosed by health professionals during treadmill running.

2.5.6 Conclusions

The controversy surrounding treadmill and overground running may be partly attributed to previous methodological designs including footwear and foot type which may have confounded the results. This experiment attempted to control for these possible confounding variables by completing barefoot running trials while ensuring all participants demonstrated a subtalar neutral foot type. As a result, this study may better demonstrate the accuracy of treadmills in representing the lower extremity kinematics associated with overground running.

This experiment demonstrated that treadmills do not alter typical lower extremity kinematics associated with overground running. In particular, there were no significant differences between treadmill and overground running with respect to rate of rearfoot angle, maximum internal tibial rotation angle and rate of internal tibial rotation. Since treadmills allow for the analysis of more gait cycles within a laboratory, they may produce more accurate kinematic results when compared to overground running. It is important to note that maximum rearfoot angle during treadmill running was significantly higher than overground running, albeit less than one degree. Although it is currently unknown whether this small difference is of clinical significance, careful interpretation should be employed when examining the magnitude of the maximum rearfoot angle obtained during treadmill running.

The next chapter will examine the mechanics of treadmill running among individuals with functional flatfoot (FFF). In addition, an investigation into the effects of orthotic intervention during treadmill running will be conducted among this clinical population.

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2.7 Appendices

Appendix 2.1: Subtalar Neutral Foot Criteria

1. ACTIVE JOINT MOVEMENTS	RANGE OF MOTION	PARTICIPANTS RANGE
HIP–seated/prone position		
Flexion - knee flexed	110° to 120°	
Extension	10° to 15°	
Abduction	30° to 50°	
Adduction	30°	
Lateral Rotation	40° to 60°	
Medial Rotation	30° to 40°	
KNEE		
Flexion	0° to 135°	
Extension	0° to 15°	
Medial Rotation of tibia on femur (knee flexed at 90°)	20° to 30°	
Lateral Rotation of tibia on femur (knee flexed at 90°)	30° to 40°	
ANKLE		
Plantarflexion	30° to 50°	
Dorsiflexion	20°	
Pronation	15° to 30°	
Supination	45° to 60°	
Metatarsal Joint Flexibility	Hold heel & twist forefoot	Good / Poor
1st MTP JOINT		
Dorsiflexion	45° to 60°	
Plantarflexion	10° to 20°	

2. STATIC ALIGNMENT	VISUAL INSPECTION	PARTICIPANT'S ALIGNMENT
HIP		
ASIS	Level?	Yes / No
PSIS	Level?	Yes / No
Iliac Crest	Level?	Yes / No
Skin fold on back	Excessive?	Yes / No
Curvature in spine	Excessive?	Yes / No
KNEE		
Genu varum - Knees apart? Genu valgum - Ankles apart?	Stand with knees and ankles together. To participate in the study both ankles and knees must touch.	Knees Touch? Yes / No Ankles Touch? Yes / No
Genu recurvatum - Hyperextension of knee in standing	Stand with knees locked. Is there excessive extension at the knee?	Yes / No
FOOT		
Weight bearing alignment		
1) Heel – tib/cal angle	Neutral calcaneal valgum is between 4^{o} and 6^{o} (Posterior view)	Angle:_______
2) Midfoot – subtalar neutral	Align tubercles of navicular and talus (Anterior view)	Is there excessive movement at the subtalar in order to align tubercles? Yes / No
3) Forefoot	Aligned with hindfoot?	Yes / No
ARCHES		
Weight bearing	Arches on ground? Neutral =Medial up, lateral down; Pronator=both down;	Medial Arch: Up / Down Lateral Arch:

	Supinator=both up	Up / Down
3) DYNAMIC ALIGNMENT	VISUAL INSPECTION	PARTICIPANT'S ALIGNMENT
Heel Toe Running		Yes / No
Symmetry between left and right	Walking Running	Yes / No Yes / No
Heel Strike	Lateral (supinated)	Yes / No
Midstance	Average pronation	Yes / No
Toe-off	Re-supination	Yes / No

Appendix 2.2: Screening Questionnaire

VOLUNTEER EXCLUSION CRITERIA **Date: (MM/DD/YYYY): ___, ___, ______**

Name: ____

Address: ______________________________________

____

City, Province: ______________, _____ Postal Code ______________

Tel #: (____)-__________ Best time to call: __________

Age: ____ yrs. Height: ______ cm Weight: _______ kg

Gender: M __ F __

Do you have any conditions that limit the use of your legs? Yes / No

If yes, how much does the condition interfere with your activities?

little or none	moderate	a great deal
€	€	€

Describe: ____

____

____

Do you have or have you ever had: **Yes / No**

		Yes / No
a)	paralysis	___
b)	epilepsy	___
c)	cerebral palsy	___
d)	multiple sclerosis	___
e)	Parkinson's disease	___
f)	stroke	___
g)	any other neurological disorder	____________________
h)	diabetes	___

Have you ever had frostbite in the lower extremities? **Yes / No**

How much do the conditions that you indicated with a 'yes' below interfere with your activities?

	Do you have or have you ever had :	**Yes / No**	little or none	moderate	a great deal
a)	problems with your heart or lungs	___	€	€	€
b)	high blood pressure	___	€	€	€
c)	blood circulation problems (generally)	___	€	€	€
	(specifically lower extremities)	___	€	€	€
d)	cancer	___	€	€	€
e)	arthritis	___	€	€	€
f)	rheumatism	___	€	€	€
g)	back problems	___	€	€	€
h)	a joint disorder	___	€	€	€
i)	a muscle disorder	___	€	€	€
j)	a bone disorder	___	€	€	€
k)	spina bifida	___	€	€	€

How much do the conditions that you indicated with a 'yes' below interfere with your activities?

	Do you have or have you ever had these <u>foot</u> problems:	**Yes / No**	little or none	moderate	a great deal
a)	bunions (hallux valgus)	___	€	€	€
b)	hammer toes	___	€	€	€
c)	calluses	___	€	€	€
d)	ulcerations	___	€	€	€
e)	plantar fasciitis	___	€	€	€
f)	any other foot problems (diagnosed or not)				
	____________________	___	€	€	€
	____________________	___	€	€	€
	____________________	___	€	€	€
	Have you ever severely injured or had surgery on your (specify)				
a)	ankle	___	€	€	€
b)	knee	___	€	€	€

c) hip		___	€	€	€
Have you ever broken any bones?		___	€	€	€

Which ones? : ____________________________________

Have you had any recent (specify)

a)	illnesses	___	€	€	€
b)	injuries	___	€	€	€
c)	operations	___	€	€	€

Do you have difficulties performing any daily activities?	___	€	€	€

Which activities?: ____________________________________

Are you currently taking any medications (prescription or over-the-counter), or other drugs?

Medication	Ailment	Frequency of use
______________	______________	______________
______________	______________	______________
______________	______________	______________

CHAPTER 3: EFFECTS OF ORTHOTIC INTERVENTION ON LOWER EXTREMITY KINEMATICS AMONG INDIVIDUALS WITH FUNCTIONAL FLATFOOT DURING TREADMILL RUNNING

3.1 Abstract

Foot orthotics are commonly prescribed to runners with functional flatfoot (FFF) with the goal of restoring the medial longitudinal arch of the foot. However the effectiveness and the mechanism by which orthotics function remains unclear in the literature (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). The objective of this study was twofold: 1) To determine if FFF is associated with excessive motion of the lower extremity during running when compared to individuals with a subtalar neutral foot type and 2) To determine if foot orthotics effectively reduce lower extremity motion during running among individuals with FFF. A total of 19 healthy subjects with FFF performed a total of 24 treadmill running trials. Participants were casted in a subtalar neutral position by a Certified Pedorthist from which medial arch supports were constructed for each participant at different percentages of their maximum medial arch height. Randomized running trials were completed at 2 different speeds (2.0m/s and 3.0m/s) under 4 conditions (barefoot and 33%, 66% and 100% of their maximum arch height). Measures of maximum rearfoot motion and maximum internal tibial rotation were used to indicate if the medial arch supports were successful at decreasing lower extremity motion. The results from this study indicate that individuals with FFF do not experience a significant increase in lower extremity motion during running when compared to individuals with a subtalar neutral foot type. In fact, the subtalar neutral group demonstrated significantly higher maximum internal tibial rotation

angles during running when compared to the FFF group. In addition, the results from this study demonstrate that orthotic intervention had a significant effect on lower extremity angles among individuals with FFF during running. In particular, maximum rearfoot angle and maximum internal tibial rotation angle decreased as the height of the medial arch support increased. Orthotic intervention did not appear to significantly affect either the rate of rearfoot angle or the rate of internal tibial rotation during running. Therefore, orthotic intervention may have a mechanical effect on the motion of the lower extremity during running. However, the extent and applicability of this effect should be further examined.

3.2 Introduction

The number of individuals participating in running as a recreational activity has increased dramatically over the past few decades (McClay, 2000). Consequently, rehabilitation centers have seen an increase in running related injuries. Previous research suggests that runners may be at a higher risk of developing an injury when compared to walkers due to the increase in maximum rearfoot angle (Perry and Lafortune, 1995) and the faster rate of pronation (Subotnick, 1985) that is observed during running. It has been proposed that near perfect biomechanics are required in order to run long distances (Subotnick, 1985) therefore it is not surprising that runners who present with abnormal biomechanics may be at an even higher risk of developing injury.

Abnormal running mechanics have been observed among individuals with functional flatfoot (FFF). These individuals characteristically present with a complete loss or a significant reduction of the medial longitudinal arch while weight bearing (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005). In an attempt to quantify the different running mechanics seen among this population researchers have found that individuals with FFF excessively and abnormally pronate for a longer period of time during the stance phase (Lee, Vabore, Thomas, Catanzariti, Kogler, Kravitz, Miller and Couture Gassen, 2005) or even throughout the entire stance phase (McClay and Manal, 1998). In addition, excessive pronators have demonstrated a significant increase in peak rearfoot eversion (McClay and Manal, 1997; McClay and Manal, 1998) and peak tibial internal rotation excursion (McClay and Manal, 1997) during running when compared to normal pronators. Further, excessive motion of the lower extremity during running has been associated with an increase in injury (McClay and Manal, 1998). However there is some controversy surrounding the literature in this area as other research has shown no significant differences in rearfoot motion between excessive and normal pronators during walking (Hunt and Smith, 2004). Thus, further research is required to better understand the mechanics that are occurring among the FFF population during running and walking.

Currently, orthotic prescription is recommended for individuals with symptomatic FFF in order to control the excessive motion of the lower extremity and thus theoretically decrease running related injuries. There is general agreement among the literature with respect to the clinical effectiveness of orthotic intervention among runners. In particular, the use of foot orthotics has been positively associated with patient satisfaction (Donatelli, Hurlbert, Conaway and St.Pierre, 1988; Moraros and Hodge, 1993) and pain reduction (Gross, Davlin and Evanski, 1991; Moraros and Hodge, 1993; Nawoczenski, Cook and Saltzman, 1995; Walter, Ng and Stoltz, 2004).

From a mechanical perspective, orthotics are thought to function by aligning the structures of the lower extremity and thus, bringing the running mechanics to resemble a more 'normal' pattern. Research pertaining to the mechanical effectiveness of orthotics has focused on measurements of rearfoot (Mundermann, Nigg, Humble and Stefanyshyn, 2003; Nester, van der Linden and Bowker, 2003; MacLean, McClay Davis and Hamill, 2006) and tibial kinematics (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Mundermann, Nigg, Humble and Stefanyshyn, 2003). Many studies suggest a significant reduction in maximum rearfoot eversion angle and velocity (Mundermann, Nigg, Humble and Stefanyshyn, 2003; MacLean, McClay Davis and Hamill, 2006) and maximum tibial rotation angle and velocity (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Mundermann, Nigg, Humble and Stefanyshyn, 2003) during running while wearing foot orthotics. However, it seems that for every study showing a positive effect of foot orthotics there is a study indicating no significant effects of foot orthotics on rearfoot motion (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Stackhouse, McClay Davis and Hamill, 2004) or tibial rotation (Nigg, Khan, Fisher and Stefanyshyn, 1998) during running.

The variability within the results may be partly attributed to study design and participant selection criteria. In terms of study design, the decision to wear footwear during the running trials may contribute to the contradictory results seen in the literature. Studies investigating the effects of footwear on lower extremity kinematics during running indicate that the heel and the heel counter of the shoe may move at different rates. More specifically, the heel has been shown to move less and at a slower rate than the heel counter (Stacoff, Reinschmidt and Stussi, 1992). Nigg (Nigg, 1986) demonstrated that a difference of 2–3 deg is observed depending on if the markers are placed on the heel or the heel counter. As a result, studies with tracking markers placed on the heel counter of the shoe may be reporting higher heel motion than what is actually occurring. In addition, footwear has been shown to significantly restrict the torsion of the foot during running, resulting in an increase in foot pronation when compared to barefoot running (Stacoff, Kaelin, Stuessi and Segesser, 1989; Stacoff, Kalin and Stussi, 1991). Therefore, controlling for footwear is essential in order to understand the mechanisms behind orthotic interventions.

Inadequate participant selection criteria may be another factor contributing to the inconclusive results of orthotic intervention studies. Many of these studies have been conducted on 'healthy' populations who do not fit the criteria for requiring foot orthotics. They do not demonstrate abnormal mechanics during running nor do they report any pain or injury of the lower extremity. It is possible that foot orthotics may not have any significant effects on lower extremity kinematic variables among 'healthy' populations since they already demonstrate normal lower extremity mechanics during running. Therefore understanding the mechanism of orthotic intervention among a clinical population, such as FFF, may provide a better understanding of the mechanical effects of foot orthotics during running.

3.2.1 Specific aims

The purpose of this study was to investigate the running mechanics associated with FFF and further, to determine the effects of orthotic intervention during running among this population. Current therapeutic modalities involve supporting the medial longitudinal arch which has been based on the assumption that individuals with FFF experience excessive motion of the lower extremity during running and thus, a higher incidence of running related injuries. This study may provide evidence for a mechanical effect of orthotics thereby indicating that realignment of the lower extremity produces a reduction in lower extremity motion. As a result, both clinicians and researchers can be confident in orthotic prescription and mechanism which may consequently lead to future enhancement of foot orthotics and ultimately to a better quality of life among individuals with FFF.

3.2.2 Hypotheses

H_O1: Individuals with FFF will demonstrate similar lower extremity kinematics during barefoot treadmill running when compared to individuals with a subtalar neutral foot type.

H_A1: Individuals with FFF will demonstrate an increase in motion of the lower extremity during barefoot treadmill running when compared to individuals with a subtalar neutral foot type.

H_O2: Individuals with FFF will demonstrate similar lower extremity motion during treadmill running while wearing medial arch supports and while running barefoot.

H_A2: Orthotic intervention will significantly affect lower extremity motion during treadmill running such that barefoot > 33% > 66% > 100% among individuals with FFF.

3.3 Method

3.3.1 Participants

A total of 19 healthy individuals (mean ± standard deviation: age, 23.8 ± 3.7 years; height, 173.8 ± 9.3 cm; weight, 71 ± 14.4 kg; 10 women, 9 men) with FFF participated in this study. This study was reviewed and approved by the University Research Ethics Board at Wilfrid Laurier University. Study participants, recruited from the university population, were informed of the requirements and signed consent forms prior to testing. Individuals were deemed eligible to participate in this study if they met our predetermined FFF criteria.

The FFF criteria consisted of static and dynamic visual inspections of pelvic and knee alignment, a rearfoot angle of greater than 10 deg and midfoot collapse while weight bearing (Appendix 3.1). Rearfoot angle was determined by using a goniometer to measure the angle that resulted from the intersection of two lines: one line connected the mid calf and the Achilles tendon and the second line connected the superior and inferior aspects of the calcaneus (Figure 3.1). All participants had a rearfoot angle of greater than 10 deg (range 10-15 deg) while standing, with the group mean ± standard deviation being 11.8 ± 1.3 deg. In addition, participants completed a screening questionnaire (Appendix 3.2) and were excluded from this study if they had any neurological or physical condition that affected the use of their lower extremities or if they regularly wore prescribed orthotics.

Based on our exclusion criteria, all subjects demonstrated characteristics consistent with FFF. Only one subject demonstrated genu varum (bow-legged spanning 4 finger width) which was classified as atypical knee alignment. Running observations indicated that 15 participants initially struck the ground with the lateral border of their heel while the remaining 4 subjects were mid/forefoot strikers. More than half (58%) of the subjects had never worn an orthotic, with the remaining 42% wearing prescribed orthotics for gym and sporting activities averaging a few times a week. None of the participants wore orthotics on a regular, daily basis.

Lower extremity running mechanics were also compared between previously tested individuals with a subtalar neutral foot type and the individuals with FFF in this study. Further details pertaining to the subtalar neutral group can be found in section 2.3.1.

Figure 3.1: Rearfoot angle as determined by the difference between the angle of the leg and the angle of the calcaneus (RF angle = $\theta_{leg} - \theta_{calcaneus}$).

3.3.2 Experiment set-up and data collection

The laboratory used for data collection measured 10m X 8m and was set up for the treadmill running trials as illustrated in Figure 3.2. The treadmill was positioned with the back, right corner resting on the force plate. The vertical force information allowed for the determination of heel contact with the force from the right foot having a higher magnitude than the left foot.

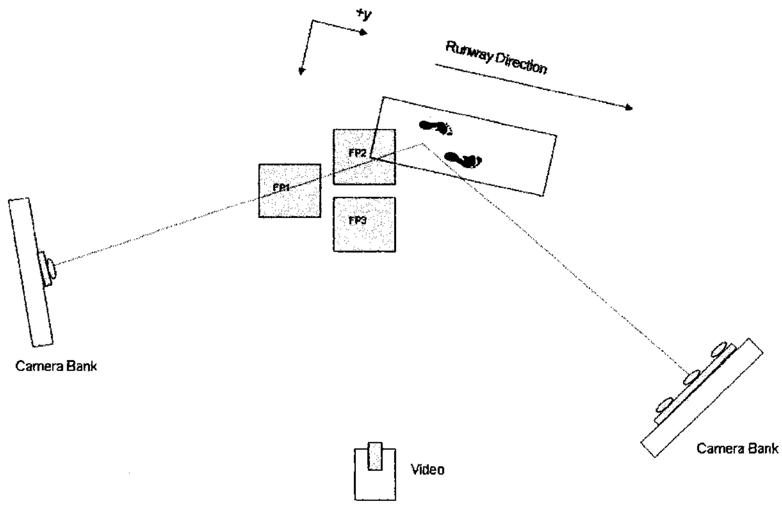

Figure 3.2: Laboratory set-up for the treadmill running trials.

Kinematic data was collected at 100Hz using 2 OptoTrak 3020 camera banks (Northern Digital Inc., Waterloo, Canada). The data from the initial 9 subjects was collected using NDI 1st Principles computer software. Due to difficulties encountered with that software, the data from the remaining 10 subjects was collected using ToolBench computer software. Since the collection software is used to record the raw marker positions, changing the software program did not affect the study results. A total of 20 infrared light-emitting diodes (IRED's) were placed on each rearfoot (4) and tibia (6) in order to track the motion of these segments (Figure 3.3).

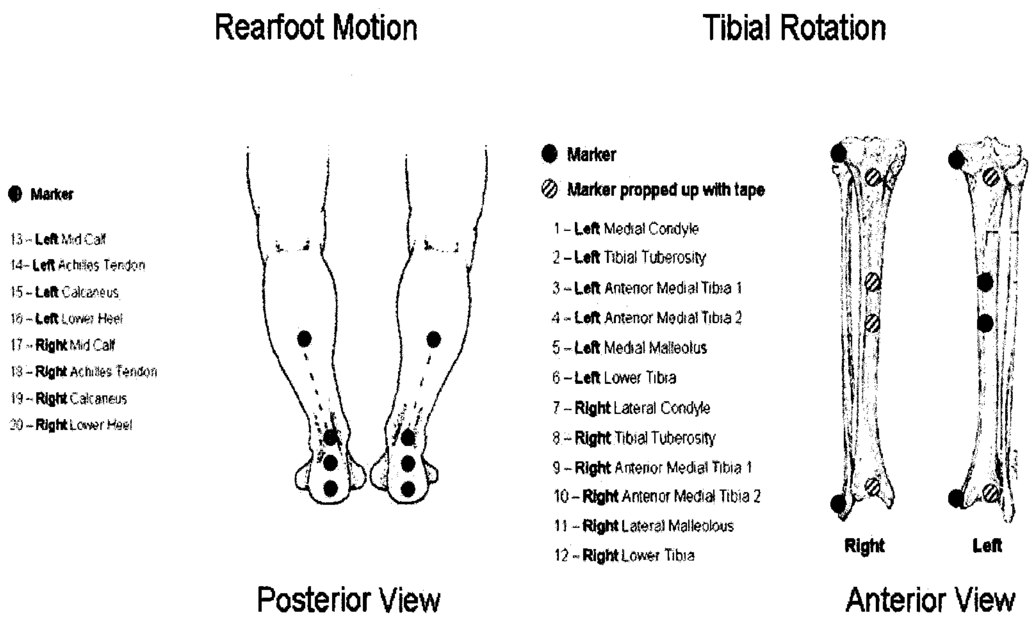

Figure 3.3: Kinematic marker placement at the rearfoot and tibia in order to track rearfoot angle and internal tibial rotation during running.

The treadmill (Figure 3.4) used for the treadmill running trials was a Precor M9.21si (Precor Inc., Bothell, WA USA). The dimensions of the running surface was 129.5cm X 43cm with the overall length, width (including handrails), and height of the treadmill measuring 170cm, 71cm and 111.8cm respectively. It contained a 2.0 hp motor.

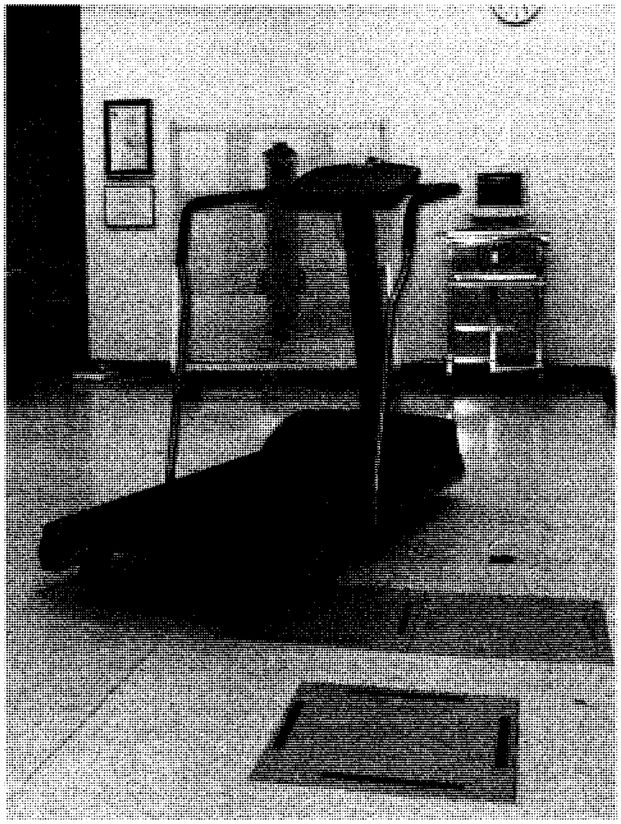

Figure 3.4: Treadmill used in this study (M9.21si, Precor Inc., Bothell, WA USA).

3.3.3 Procedure

This study consisted of three sessions. During Session 1 the FFF criteria and the exclusion questionnaire were completed. If the participants were deemed eligible based on the criteria and the questionnaire they were invited back to complete Session 2.

During Session 2, participants met with a Certified Pedorthist who made a subtalar neutral foam cast of their feet. From the foam cast a mold of the foot was made by Pedorthic Services Lab (Waterloo, Canada). Maximum medial arch height was then measured from this foot mold. Customized medial arch supports of 33%, 66% and 100% of the maximal medial arch height were created for each participant. Please refer to

Appendix 3.3 for details pertaining to the construction of the medial arch supports. The maximum medial arch heights as well as the 33%, 66% and 100% arch support heights for each participant can be seen in Table 3.1. Figure 3.5 illustrates the final arch supports of one subject for each orthotic condition.

Figure 3.5: Moving left to right, this shows the completed arch supports of 100%, 66% and 33% for one subject.

Table 3.1: Measurements of maximum medial arch height and the subsequent arch supports at 33%, 66% and 100%. All values were rounded to the nearest mm. The maximum height of the arch support was 16mm therefore the 100% condition could only have a maximum of 16mm regardless of the measurement. The calculations for 33% and 66% were completed using the measured maximum medial arch height as determined by the electronic caliper.

		Max Medial Arch Height (mm)		33% Max Medial Arch Height (mm)		66% Max Medial Arch Height (mm)		100% Max Medial Arch Height (mm)	
Subject		**Caliper Measurement**							
		Left	**Right**	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**
S02		16	16	5	5	11	11	16	16
S03		20	19	7	6	13	12	16	16
S05		11	16	4	5	7	11	11	16
S06		12	14	4	5	8	9	12	14
S08		14	12	5	4	9	8	14	12
S09		17	14	6	5	11	9	16	14
S10		16	15	5	5	11	10	16	15
S11		11	15	4	5	7	10	11	15
S12		16	16	5	5	11	11	16	16
S13		16	18	5	6	11	12	16	16
S14		20	20	7	7	13	13	16	16
S15		13	16	4	5	9	11	13	16
S16		14	20	5	7	9	13	14	16
S17		23	19	8	6	15	13	16	16
S18		14	17	5	6	9	11	14	16
S19		15	15	5	5	10	10	15	15
S20		16	17	5	6	11	11	16	16
S21		19	17	6	6	13	11	16	16
S22		14	17	5	6	9	11	14	16

Once the arch supports had been constructed, participants were invited back for the Session 3. This session consisted of treadmill running at two different speeds under 4 orthotic conditions. Therefore a total of 8 randomized conditions were completed including:

Barefoot - 2.0m/s
Barefoot - 3.0m/s
Medial Arch Support (33%) - 2.0m/s
Medial Arch Support (33%) -3.0m/s
Medial Arch Support (66%) - 2.0m/s
Medial Arch Support (66%) - 3.0m/s
Medial Arch Support (100%) - 2.0m/s
Medial Arch Support (100%) - 3.0m/s

Prior to beginning the treadmill running trials, the kinematic markers were applied to the rearfoot and tibia of each leg on the participant as illustrated in Figure 3.3. The rearfoot angle was measured and recorded with the subject standing barefoot prior to commencing the study. They were then given as many treadmill practice trials as required in order to feel comfortable running at 2.0m/s and 3.0m/s. A spotter stood beside them throughout all running trials to ensure their safety. Once participants felt comfortable on the treadmill they stepped off in order to have the arch support secured to their foot. Athletic tape was used to adhere the arch supports to the plantar surface of the foot, specifically, to the medial longitudinal arch. Tape strips of 15-20cm in length were placed on the medial aspect of the foot and wrapped under the bottom of the foot covering the arch support before being secured to the lateral aspect of the foot. A smaller strip of athletic tape was used to anchor the ends of the tape to the foot ensuring not to cover the metatarsal phalangeal (MTP) joints. This attachment technique sufficiently kept the arch support in place throughout testing. In addition, participants reported that the taping did not limit the normal movement of their foot.

Once the participant was back on the treadmill the speed was increased until the desired speed was achieved. A total of three 5 second trials were collected before the treadmill speed was returned to a comfortable walking pace as set by the participant. On the participants command, the treadmill speed was again increased to record three 5 second consecutive trials at the remaining speed. Once these trials were collected the speed was decreased to a comfortable walking pace until the participant indicated that they had sufficiently cooled down and were ready to step off. The next arch support was then adhered to the foot and the rearfoot angle was measured prior to beginning the running trials. Rearfoot measurements were taken during each orthotic condition to ensure that the arch supports were having an effect on the structure of the medial longitudinal arch. In addition, the participant was asked to report which orthotic condition was the most and least comfortable during running. Following completion of all treadmill running trials, the participant stood quietly on the treadmill while a 5 second static trial was recorded. Thus, Session 3 involved the completion of 25 trials (24 running and 1 static).

3.3.4 Data analysis and statistics

The 8 conditions in this protocol allowed for the analysis of orthotic intervention on lower extremity kinematics during treadmill running at two different speeds. The independent variables in this analysis were orthotic intervention (barefoot, 33%, 66%, or 100%), velocity (2.0m/s or 3.0m/s) and foot (right or left). The barefoot running conditions from this FFF population (n=19) were also compared to barefoot running trials from the subtalar neutral population (n=19) in order to determine if individuals with FFF experienced excessive rearfoot motion and internal tibial rotation during running (Table 3.2). The independent variables in this analysis were group (FFF or subtalar neutral), velocity (2.0m/s or 3.0m/s) and foot (right or left). The dependent kinematic variables in both analyses were rearfoot angle (used to measure foot pronation) and internal tibial rotation. Please refer to Table 3.3 for a description of each kinematic variable analyzed in this study. A priori analysis of estimation of sample size was conducted prior to subject recruitment in order to ensure that a statistical power of 0.8 was achieved.

Table 3.2: Repeated measures analysis of variance was conducted between subtalar neutral and FFF populations with no arch support as well as between orthotic interventions among the FFF populations.

	SUBTALAR NEUTRAL		FFF	
No Arch	Rearfoot Motion 2.0m/s & 3.0m/s	Tibial Rotation 2.0m/s & 3.0m/s	Rearfoot Motion 2.0m/s & 3.0m/s	Tibial Rotation 2.0m/s & 3.0m/s
33% Arch			Rearfoot Motion 2.0m/s & 3.0m/s	Tibial Rotation 2.0m/s & 3.0m/s
66% Arch			Rearfoot Motion 2.0m/s & 3.0m/s	Tibial Rotation 2.0m/s & 3.0m/s
100% Arch			Rearfoot Motion 2.0m/s & 3.0m/s	Tibial Rotation 2.0m/s & 3.0m/s

Rearfoot angle was used as an estimate of foot pronation due to the difficulty in measuring the tri-planar movement this variable. In this study, rearfoot angle was calculated using the four rearfoot kinematic markers (Figure 3.3). The top two markers allowed for the determination of the absolute angle of the leg relative to the horizontal; the bottom two markers were used to determine the absolute angle of the calcaneus relative to the horizontal. The relative angle of the rearfoot was determined by subtracting the absolute angle of the leg from the absolute angle of the calcaneus:

$$\text{Rearfoot angle} = \theta_{leg} - \theta_{calcaneus}$$

The second dependent variable, internal tibial rotation, was tracked by six kinematic markers that were placed on the tibia (Figure 3.3). Computer software (Visual 3D) was used to create a model of the right and left tibia for each subject. This model was created from a standing trial and then assigned to the running trials. Thus, this study analyzed internal tibial rotation relative to the static position of the tibia. Internal tibial rotation was calculated within the Visual 3D software and defined as rotation around the vertical (z) axis in the transverse plane. Both rearfoot motion and internal tibial rotation were calculated for every frame within each 5 second trial.

Table 3.3: A description of the kinematic variables analyzed in this study. All angles and rates were calculated for every frame during each 5 second trial.

Kinematic Variable	Description
Maximum Rearfoot Angle	Rearfoot angle (RF_θ) was determined by calculating the difference between the leg and calcaneal angles. Maximum rearfoot angle (Max RF_θ) was defined as the maximum rearfoot angle achieved during the stance phase minus the rearfoot angle at heel contact: $\text{Max } RF_\theta = RF_{\theta \text{ max stance}} - RF_{\theta \text{ heel contact}}$
Rate of Rearfoot Angle	Rate of rearfoot angle (RF_{rate}) was defined as the rate at which the rearfoot achieved the maximum rearfoot angle during stance. It was calculated by dividing the Max RF_θ by the difference in time from heel contact to maximal stance: $RF_{rate} = \dfrac{\text{Max } RF_\theta}{t_{RF\theta \text{ max stance}} - t_{RF\theta \text{ heel contact}}}$.
Maximum Internal Tibial Rotation Angle	Internal tibial rotation angle (ITR_θ) was defined as rotation around the vertical (z) axis in the transverse plane and was calculated using Visual 3D software. Maximum ITR_θ was defined as the difference between ITRθ at heel contact and maximal stance:

	$Max\ ITR_\theta = ITR_{\theta\ max\ stance} - ITR_{\theta\ heel\ contact}$
Rate of Internal Tibial Rotation	Rate of internal tibial rotation (ITR_{rate}) was defined as the rate at which the tibia achieved the maximum internal rotation angle during stance. It was calculated by dividing the Max ITR_θ by the difference in time from heel contact to maximal stance : $ITR_{rate} = \frac{Max\ ITR_\theta}{t_{ITR\theta\ max\ stance} - t_{ITR\theta\ heel\ contact}}$.
Foot Placement (Medial-lateral)	The medial-lateral foot placement during treadmill running was determined using the x component of markers 16 or 20 for the left and right foot, respectively.
Foot Placement (Anterior-posterior)	The anterior-posterior foot placement during treadmill running was determined using the y component of markers 16 or 20 for the left and right foot, respectively.

A program was written in Visual Basic specifically for the analyses in this study. This program allowed for the selection of OptoTrak, force plate and Visual 3D data and then displayed them in a graph. This graph showed the position of the right (marker 20) or left (marker 16) foot as well as the vertical force. The vertical force data was collected during this study in order to more accurately determine when heel contact and toe off had occurred. After selecting an area of the graph that correlated with an increase in vertical force, rearfoot and internal tibial rotation curves were produced. From these curves, maximum angles and rates of both the rearfoot and tibia were determined. During each trial, 5 stance phases were selected and analyzed for each of the right and left foot.

The results were analyzed using the SAS computerized statistical package. Analysis of the orthotic intervention was employed using a three factor (4 conditions X 2 speeds X 2 feet) within-subject repeated measures ANOVA with the a priori alpha set at 0.05. In order to determine where the significant differences found in the ANOVA's occurred, Tukey's Studentized Range (HSD) post hoc procedure was employed. Analysis of foot type on lower extremity kinematics was completed using a one factor (2 groups (foot type)) between subjects ANOVA with the a priori alpha set at 0.05. All rearfoot and tibial rotation outliers with a standard residual greater than 3.5 were investigated. A total of 70 tibial rotation outliers were excluded from this study for reasons that included missing data during stance ranging from 15 frames to the entire stance phase, externally rotated tibia at heel contact or maximum internal rotation of the tibia prior to heel contact.

3.4 Results

The purpose of this study was twofold: 1) to determine if individuals with FFF experienced an increase in lower extremity motion during running when compared to individuals with a subtalar neutral foot type and 2) to determine the effects of medial arch supports on lower extremity motion during running among individuals with FFF. For the purpose of this study, the medial arch support conditions are referred to as orthotic intervention.

This section begins with a comparison of the lower extremity kinematics during running among individuals with subtalar neutral and FFF foot types. Specifically, analysis between the 2 groups in terms of the demographics, rearfoot motion and internal tibial rotation variables are discussed. Then, the effects of orthotic intervention among individuals with FFF during running are presented. First, a description of the effects of medial arch supports on static rearfoot angle as well as the participants perceived comfort ratings for each level of medial arch support are presented. Then, the results for each dependent variable are presented: 1) maximum rearfoot angle, 2) rate of rearfoot angle, 3) maximum internal tibial rotation angle, 4) rate of internal tibial rotation, and 5) foot placement during treadmill running. The result of each variable begins with a description of how orthotic intervention, running speed and foot affected that particular variable as well as any interactions that occurred. The included figures illustrate the effects of orthotic intervention and running speed on the dependent variable of interest. Following these figures are graphs which illustrate the interactions for each variable if they were found to occur. The tables at the end of this section summarize the effects of orthotic intervention (Table 3.6), running speed (Table 3.7) and foot (Table 3.8) on each of the dependent variables.

3.4.1 Lower extremity kinematics during running between subtalar neutral and FFF foot types

Demographics

There were no significant differences between the participants within the two groups with respect to age (p=0.0646), weight (p=0.9316) or height (p=0.4718). In addition, both groups were comprised of 10 females and 9 males. As expected, the individuals with FFF had a significantly higher static rearfoot angle when compared to individuals with a subtalar neutral foot type for both the left (p<0.0001) and right (p<0.0001) feet. Please refer to table 3.4 for further details pertaining to the characteristics of these two groups. Although a comparison cannot be made between groups, the FFF group had an average medial arch height of 15.6 ± 3.2mm (range: 11-23mm) and 16.5 ± 2.1mm (range: 12-20mm) for the left and right foot, respectively (Table 3.1).

Table 3.4: Characteristics of the two groups: Subtalar neutral and FFF. Values are presented as mean (standard deviation). '*' denotes significance.

Group	Age (yrs)	Weight (kg)	Height (cm)	Static Rearfoot Angle Left (deg) *	Static Rearfoot Angle Right (deg) *	Gender
Subtalar Neutral	21.8 (3.2)	70.5 (13.7)	172.1 (9.8)	4.6 (1.1)	4.4 (1.0)	10 females 9 males
FFF	23.8 (3.7)	71.0 (14.4)	173.8 (9.3)	11.5 (1.2)	12.2 (1.3)	10 females 9 males
p value	0.0646	0.9316	0.4718	<0.0001	<0.0001	

Rearfoot Motion

Maximum rearfoot angle was not significantly different between subtalar neutral and FFF individuals (p=0.0962). The maximum rearfoot angles obtained for subtalar neutral and FFF individuals were 9.7 ± 3.3 deg and 10.4 ± 4.0 deg, respectively (Figure 3.6). The rate of rearfoot angle did not differ significantly between the two groups (p=0.3478) as subtalar neutral and FFF individuals demonstrated values of 100.9 ± 46.8 deg/sec and 114.6 ± 70.0 deg/sec, respectively (Figure 3.7). Table 3.5 provides a summary of the rearfoot motion variables between the two groups.

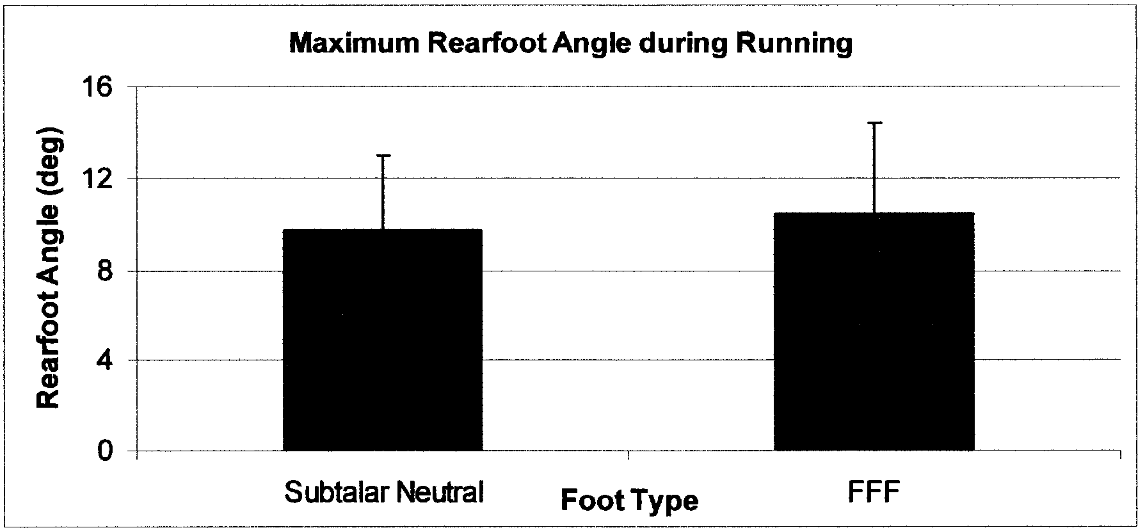

Figure 3.6: Maximum rearfoot angle (mean + standard deviation) achieved during running between the two groups.

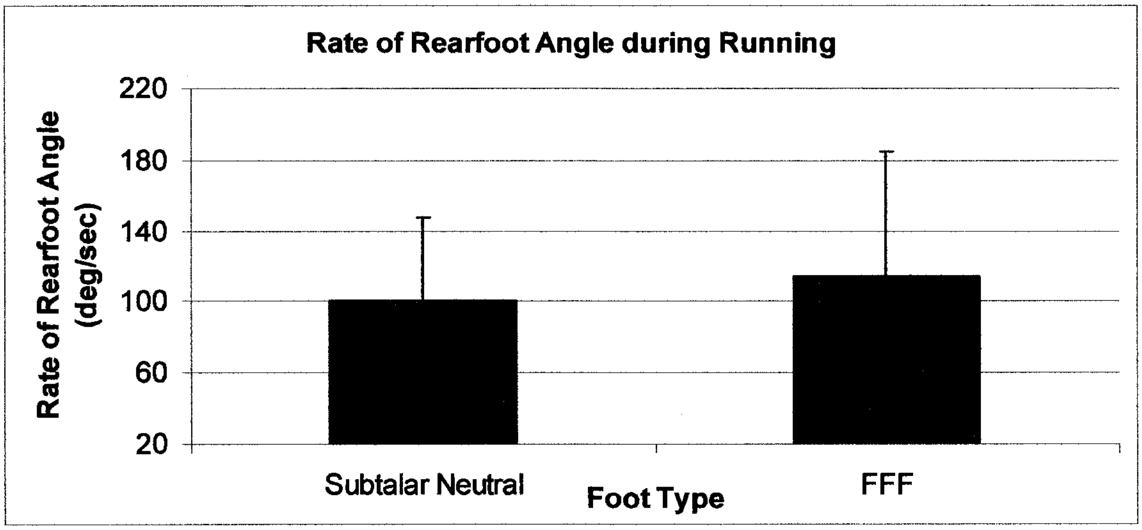

Figure 3.7: Rate of rearfoot angle (mean + standard deviation) achieved during running between the two groups.

Internal Tibial Rotation

Maximum internal tibial rotation angle was significantly higher among the subtalar neutral individuals when compared to the FFF individuals ($p<0.0001$). The subtalar neutral and FFF groups demonstrated maximum internal tibial rotation angles of 13.5 ± 8.4 deg and 10.0 ± 7.6 deg, respectively (Figure 3.8). The rate of internal tibial rotation was not significantly different between the two groups ($p=0.3885$) as subtalar neutral and FFF demonstrated values of 104.6 ± 69.0 deg/sec and 102.0 ± 80.3 deg/sec, respectively (Figure 3.9). Table 3.5 provides a summary of the internal tibial rotation variables between groups.

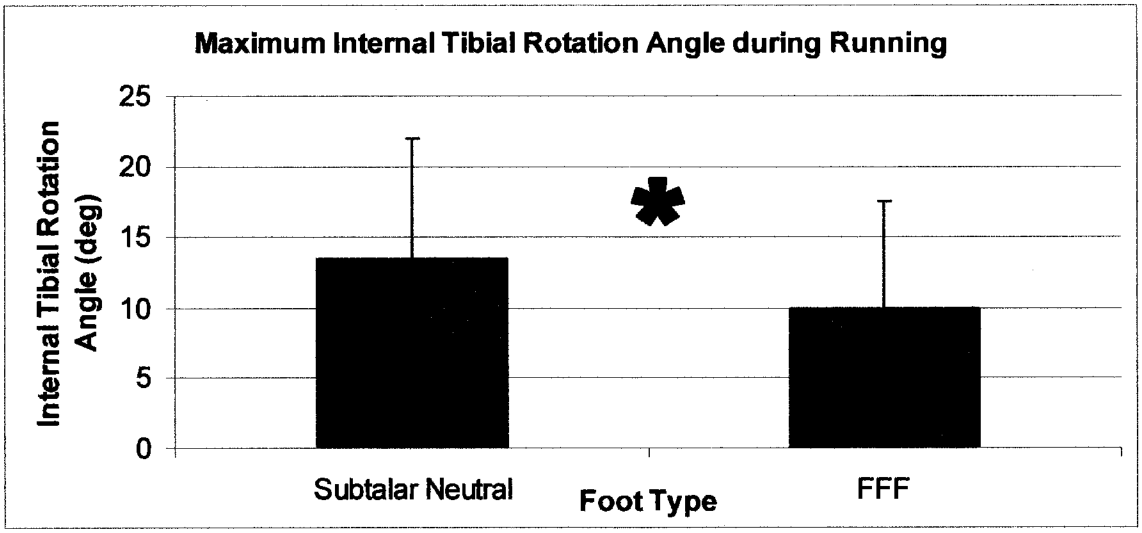

Figure 3.8: Maximum internal tibial rotation angle (mean + standard deviation) achieved during running between the two groups. The symbol '✱' denotes significance of $p<0.05$.

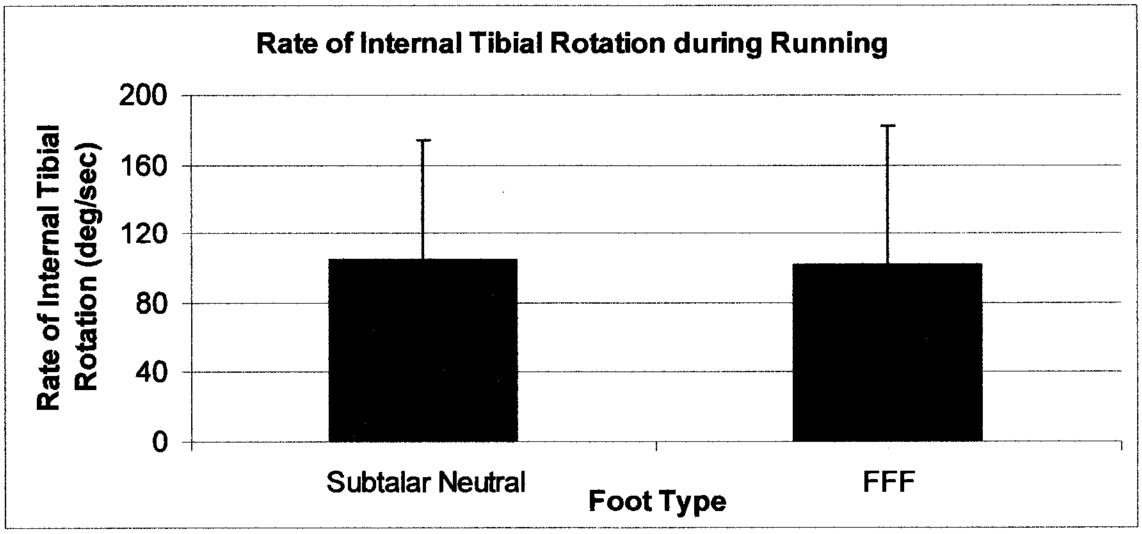

Figure 3.9: Rate of internal tibial rotation (mean + standard deviation) achieved during running between the two groups.

Table 3.5: A summary of the lower extremity kinematic variables between the two groups. '*' denotes significance of p<0.05.

Group	Maximum Rearfoot Angle (deg)	Rate of Rearfoot Angle (deg/sec)	Maximum Internal Tibial Rotation Angle (deg) *	Rate of Internal Tibial Rotation (deg/sec)
Subtalar Neutral	9.7 (3.3)	100.9 (46.8)	13.5 (8.4)	104.6 (69.0)
FFF	10.4 (4.0)	114.6 (70.0)	10.0 (7.6)	102.0 (80.3)
p value	0.0962	0.3478	<0.0001	0.3885

3.4.2 Effects of medial arch supports on static rearfoot angle

Static measurement of the rearfoot revealed a decrease in the rearfoot angle between each orthotic condition as the medial arch height increased. Static rearfoot angles were 11.8 ± 1.3 deg, 11.3 ± 1.4 deg, 10.1 ± 1.3 deg and 9.2 ± 1.3 deg for each of the BF, 33%, 66% and 100% orthotic intervention conditions, respectively (Figure 3.10). Analysis involving a single factor ANOVA revealed significance between each orthotic condition ($p<0.05$) such that the static rearfoot angle at BF > 33% > 66% > 100%. This result indicates that the medial arch supports were having a significant effect on the structure of the foot.

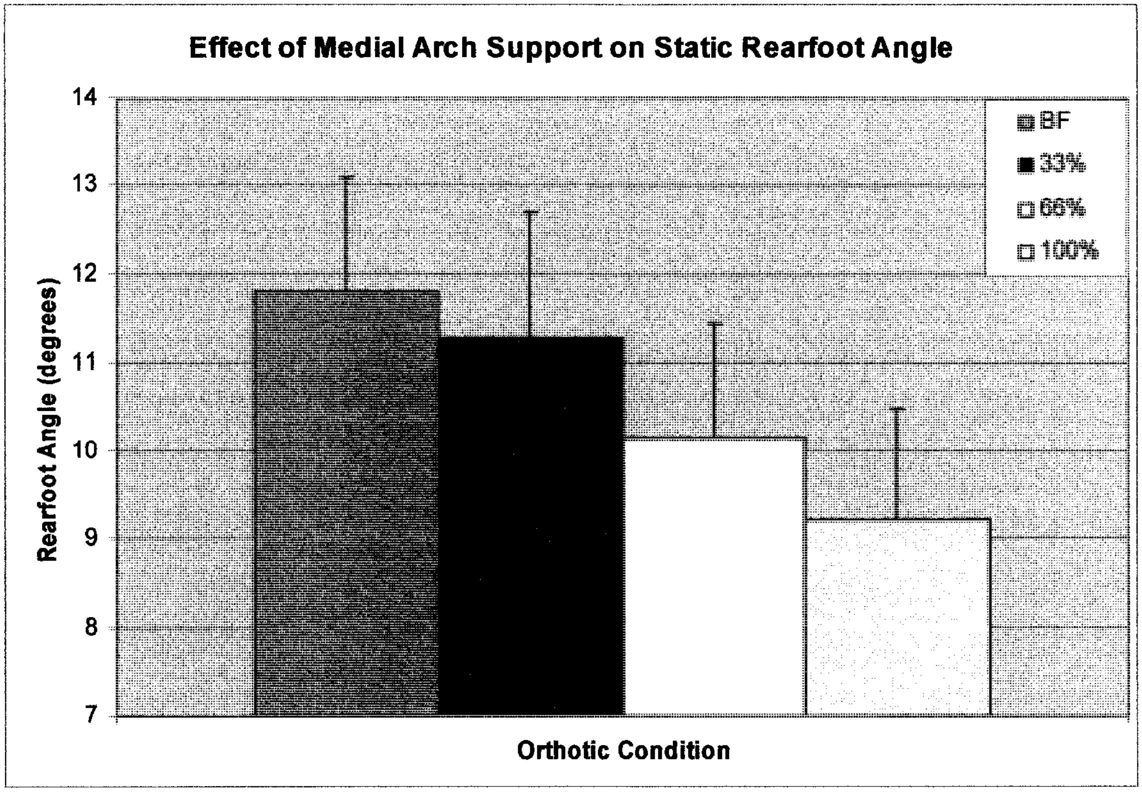

Figure 3.10: Effects of medial arch support on the static rearfoot angle measurement. Conditions include barefoot (BF) and 33%, 66% and 100% of the participants' maximum medial arch height. Significance ($p<0.05$) was achieved between each orthotic condition.

3.4.3 Perceived comfort ratings for each level of medial arch support

Half of the participants reported that the orthotic intervention condition of 33% was the most comfortable to wear. In addition, 37.5% and 12.5% reported that the most comfortable orthotic condition was 66% and 100%, respectively. None of the participants reported the barefoot condition as being the most comfortable (Figure 3.11). A chi-square analysis revealed a critical value for k-1 = 3 *df*, $\chi^2_{.05}(3) = 7.82$. Our calculations for the chi-square test demonstrated a value of 10 which is greater than the critical value of 7.82 therefore the participant's choices were not random. Participants chose the orthotic conditions of 33% and 66% as the most comfortable conditions at greater than chance levels.

The majority of the participants reported that the 100% orthotic condition was the least comfortable (68.75%). A total of 12.5% of the participants reported that each of the orthotic conditions of 33% and 66% were the least comfortable. None of the participants reported the barefoot condition as being the least comfortable (Figure 3.12). In addition, one participant reported that all the orthotic conditions were comfortable. A chi-square analysis revealed a critical value for k-1 = 3 *df*, $\chi^2_{.05}(3) = 7.82$. Our calculations for the chi-square test demonstrated a value of 19.4 which is greater than the critical value of 7.82 therefore the participant's choices were not random. Participants chose the orthotic condition of 100% as the most uncomfortable condition at greater than chance levels.

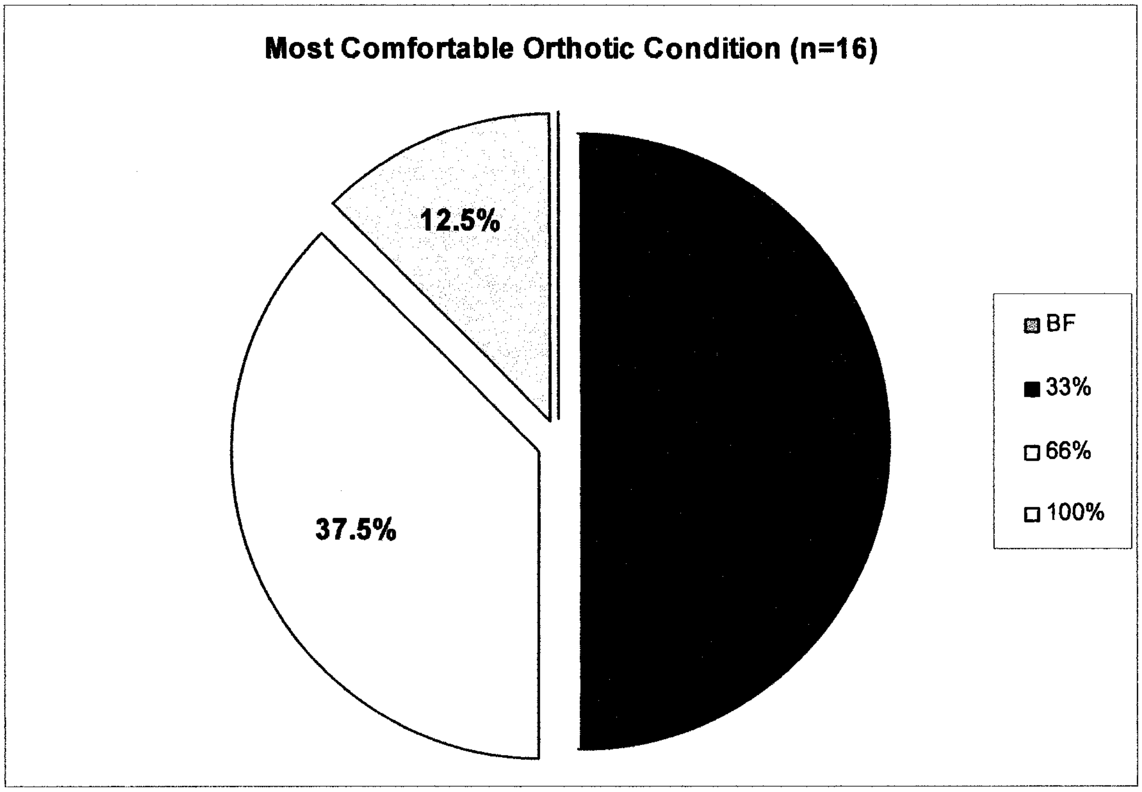

Figure 3.11: Participants perceived comfort ratings for the most comfortable orthotic condition (Barefoot, 33%, 66% and 100%).

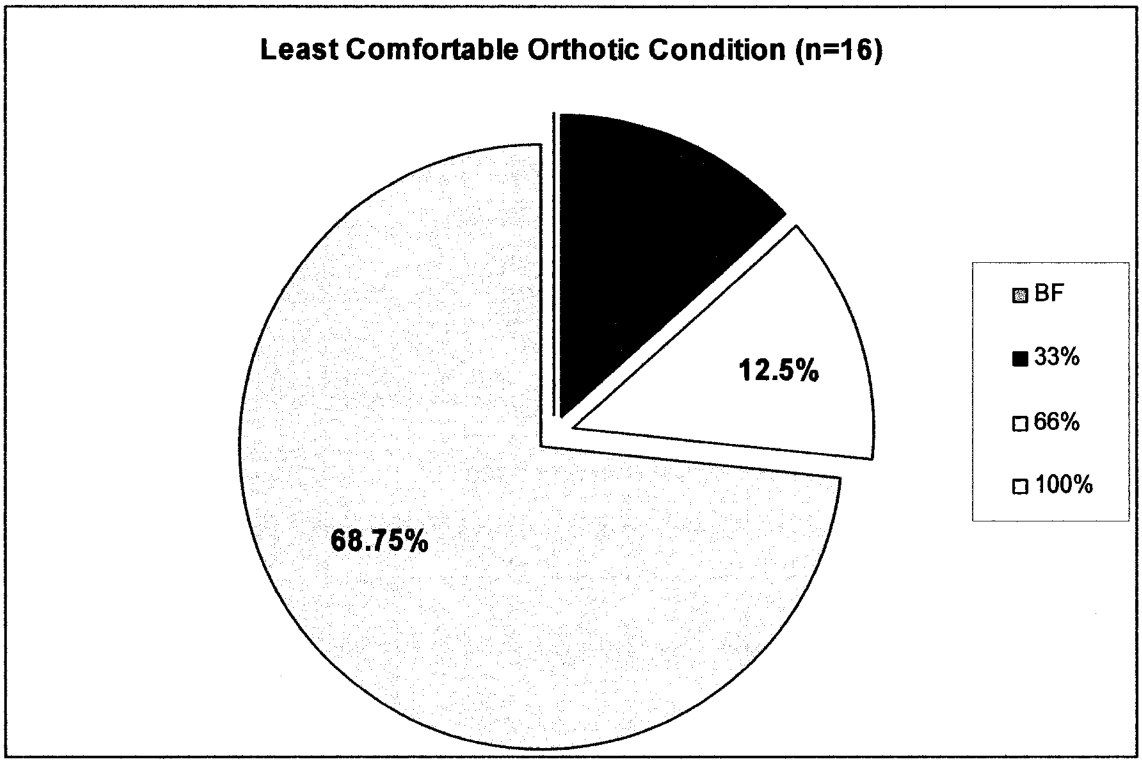

Figure 3.12: Participants perceived comfort ratings for the least comfortable orthotic condition (Barefoot, 33%, 66% and 100%).

3.4.4 Maximum rearfoot angle

Orthotic intervention had a significant effect on maximum rearfoot angle ($p<0.0001$). More specifically, the maximum rearfoot angle significantly decreased as the medial arch height increased. Significance was reached between each condition with maximum rearfoot angles of 10.4 ± 4.0 deg, 10.1 ± 3.8 deg, 9.4 ± 3.7 deg and 8.8 ± 3.6 deg for the barefoot, 33%, 66% and 100% conditions, respectively (Table 3.6). Running speed also had a significant effect on maximum rearfoot angle ($p=0.0001$) as running at 2.0m/s and 3.0m/s was associated with maximum rearfoot angles of 9.2 ± 3.7 deg and 10.2 ± 3.9 deg, respectively (Table 3.7). Maximum rearfoot angle was not significantly affected by foot ($p=0.4916$) as the left and right foot had maximum angles of 9.4 ± 3.1 deg and 9.9 ± 4.4 deg, respectively (Table 3.8). Figure 3.13 illustrates the effects of orthotic intervention and running speed on maximum rearfoot angle.

No significant interactions were found to occur with maximum rearfoot angle.

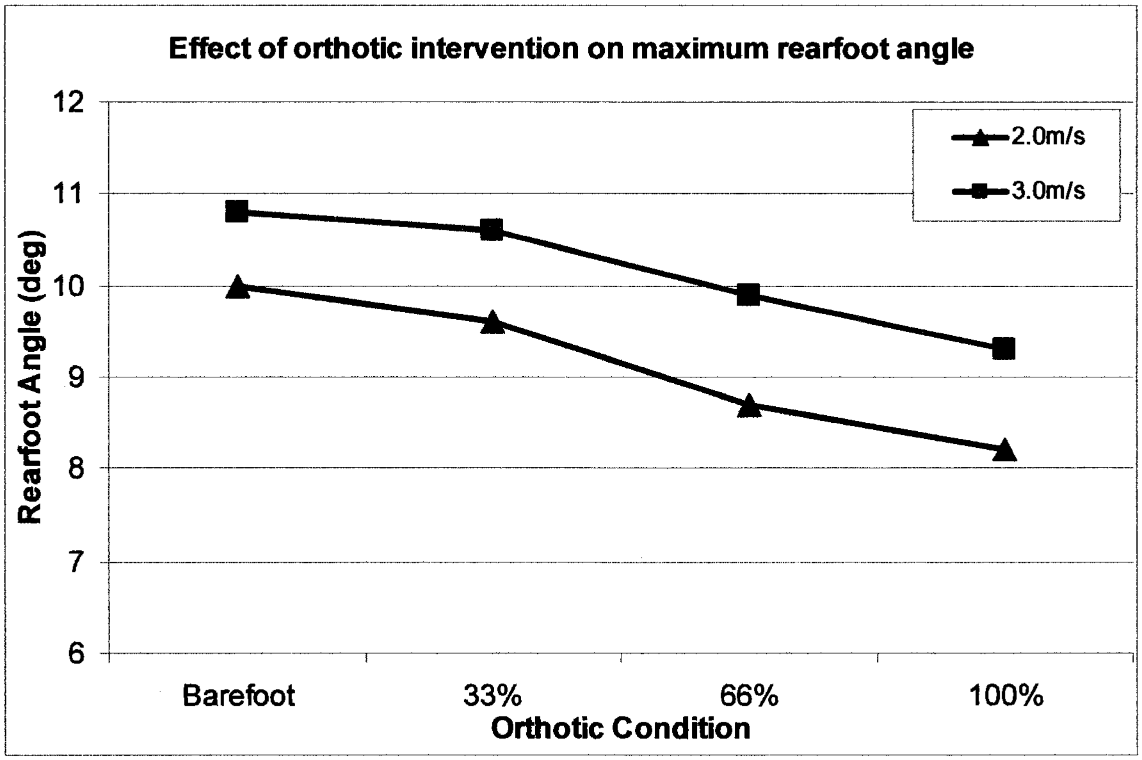

Figure 3.13: The effect of orthotic intervention on maximum rearfoot angle while running at 2.0m/s and 3.0m/s. Mean values are presented; standard deviations are presented on page 130.

3.4.5 Rate of rearfoot angle

Orthotic intervention did not have a significant effect on the rate of rearfoot angle ($p=0.2198$). The rate of rearfoot angles obtained for the BF, 33%, 66% and 100% conditions were 115.2 ± 69.6 deg/sec, 121.8 ± 76.7 deg/sec, 119.3 ± 79.1 deg/sec and 114.6 ± 79.5 deg/sec, respectively (Table 3.6). The rate of rearfoot angle was significantly affected by running speed ($p<0.0001$). Specifically, running at 2.0m/s and 3.0m/s was associated with values of 99.2 ± 61.5 deg/sec and 136.2 ± 84.8 deg/sec, respectively (Table 3.7). A significant difference with respect to the rate of rearfoot angle was observed between feet ($p<0.0045$). For example, the left foot (128.7 ± 79.7 deg/sec) had an increased rate of rearfoot angle when compared to the right foot (106.7 ± 71.3 deg/sec) (Table 3.8). Figure 3.14 illustrates the effects of orthotic intervention and running speed on the rate of rearfoot angle.

There was a significant interaction between running speed and foot ($p=0.0435$). More specifically, the rate of rearfoot angle was significantly less with the right foot (86.4 ± 56.4 deg/sec) when compared to the left foot (112.1 ± 63.8 deg/sec) while running at 2.0m/s only (Figure 3.15). A significant interaction was also demonstrated between orthotic intervention and foot ($p=0.0186$). More specifically, there was a decrease in the rate of rearfoot angle with the right foot as the medial arch height increased (BF, 110.4 ± 65.7 deg/sec; 33%, 109.1 ± 67.9 deg/sec; 66%, 110.7 ± 80.5 deg/sec; 100%, 96.6 ± 69.3 deg/sec) whereas the left foot experienced an increase in the rate of rearfoot angle as the medial arch height increased (BF, 119.9 ± 72.9 deg/sec; 33%, 134.3 ± 82.7 deg/sec; 66%, 127.8 ± 76.9 deg/sec; 100%, 132.7 ± 84.9 deg/sec) while running at 2.0m/s (Figure 3.16).

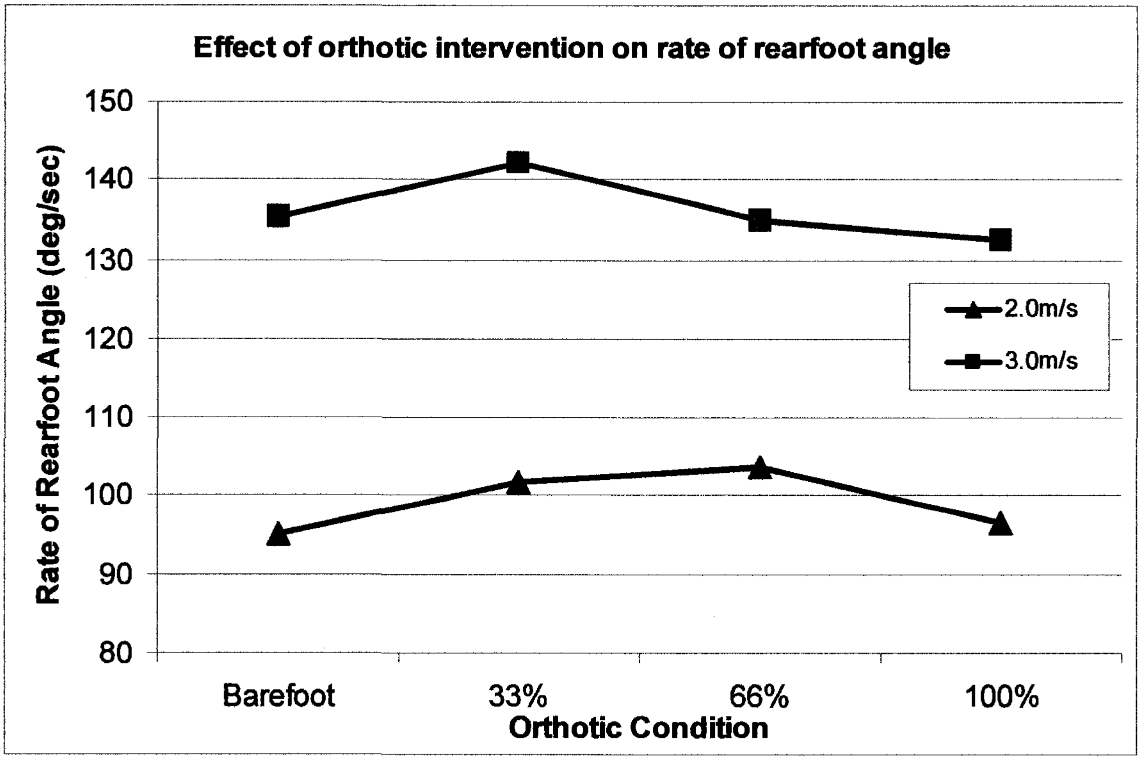

Figure 3.14: The effect of orthotic intervention on the rate of rearfoot angle while running at 2.0m/s and 3.0m/s. Mean values are presented; standard deviations are presented on page 132.

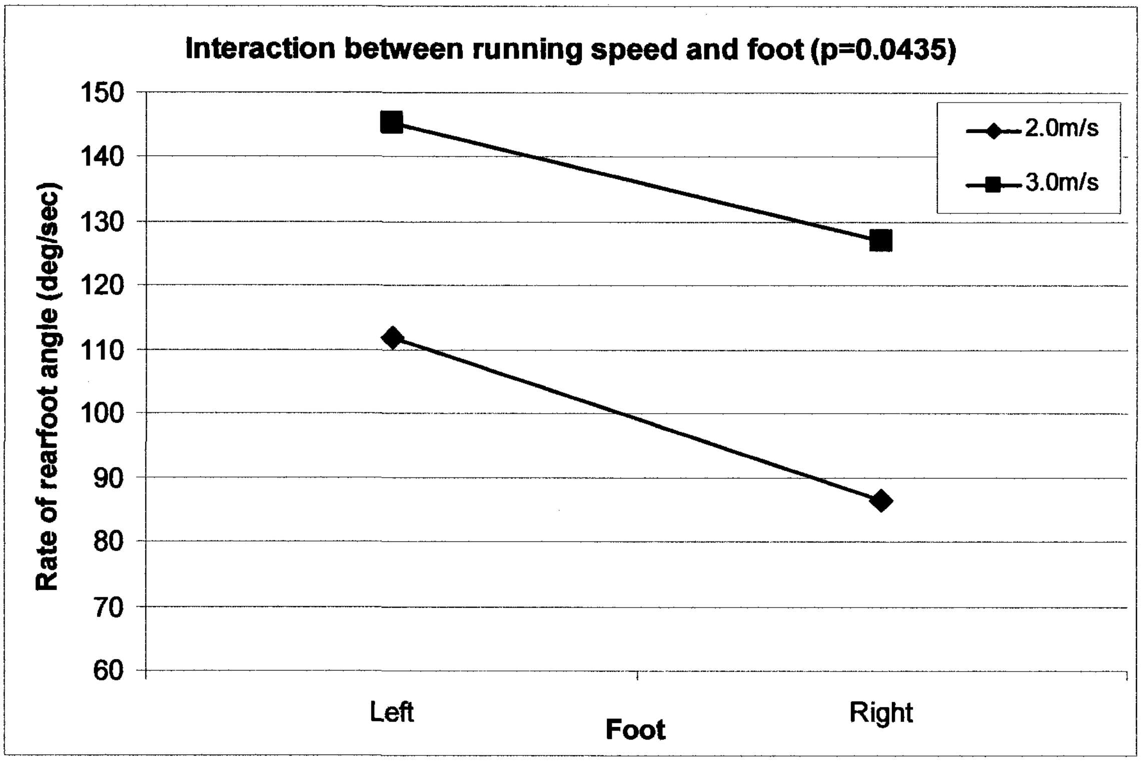

Figure 3.15: The interaction between running speed and foot (p=0.0435) on the rate of rearfoot angle. Mean values presented; standard deviations are presented on page 132.

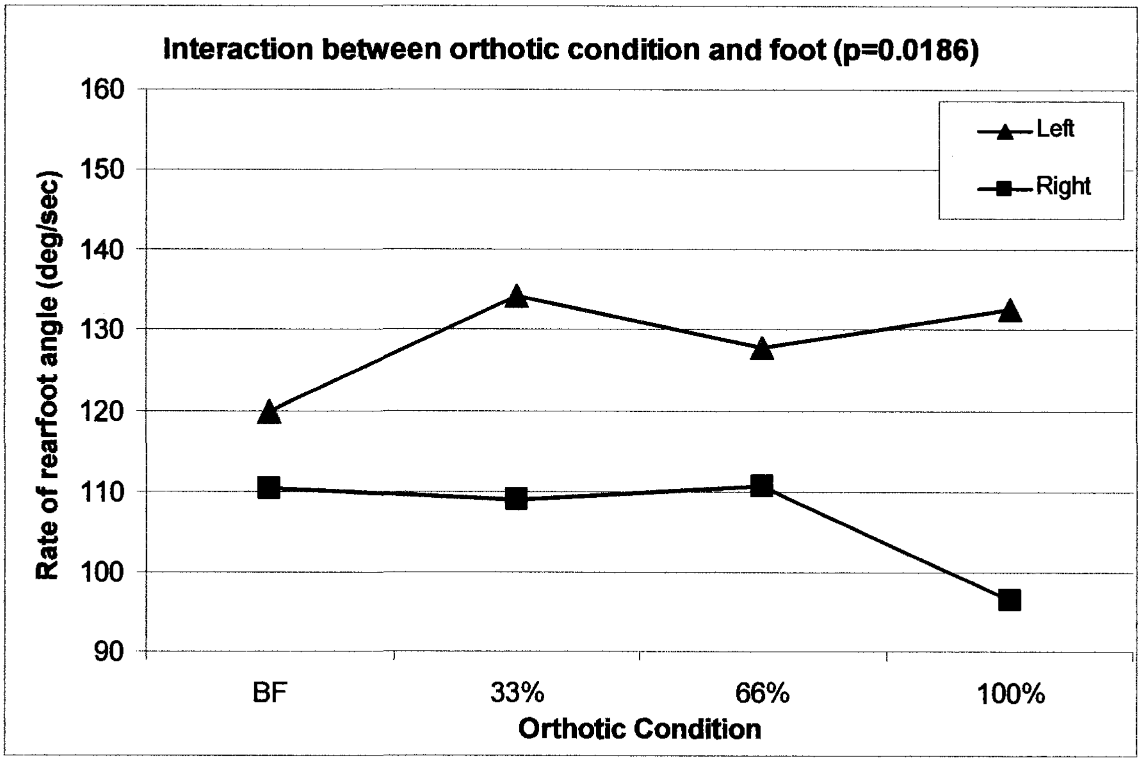

Figure 3.16: The interaction between orthotic condition and foot (p=0.0186) on the rate of rearfoot angle. Mean values presented; standard deviations are presented on page 132.

3.4.6 Maximum internal tibial rotation angle

Maximum internal tibial rotation angle was significantly affected by orthotic intervention (p=0.0019) as maximum internal tibial rotation angles of 10.0 ± 7.5 deg and 10.2 ± 8.3 deg, 9.6 ± 7.6 deg and 8.9 ± 7.2 deg were associated with BF, 33%, 66% and 100%, respectively. Significance was determined between each orthotic condition with the exception of BF vs 33% (Table 3.6). Running speed had a significant effect on maximum internal tibial rotation angles (p=0.0087) as 2.0m/s and 3.0m/s demonstrated 9.1 ± 7.5 deg and 10.3 ± 7.8 deg, respectively (Table 3.7). Maximum internal tibial rotation angle was also significantly affected by foot (p=0.0072) as the left and right foot produced maximum internal tibial rotation angles of 12.1 ± 8.2 deg and 7.2 ± 6.2 deg, respectively (Table 3.8). Figure 3.17 illustrates the effects of orthotic intervention and running speed on maximum internal tibial rotation angle.

There was a significant interaction between orthotic intervention and running speed (p=0.0035). More specifically, there was a significant decrease in maximum internal tibial rotation angle as medial arch height increased while running at 2.0m/s (BF, 9.7 ± 7.3 deg; 33%, 9.2 ± 8.0 deg; 66%, 8.8 ± 7.4 deg; 100%, 8.5 ± 7.1 deg) when compared to running at 3.0m/s (BF, 10.3 ± 7.6 deg; 33%, 11.3 ± 8.5 deg; 66%, 10.3 ± 7.7 deg; 100%, 9.2 ± 7.3 deg) (Figure 3.18).

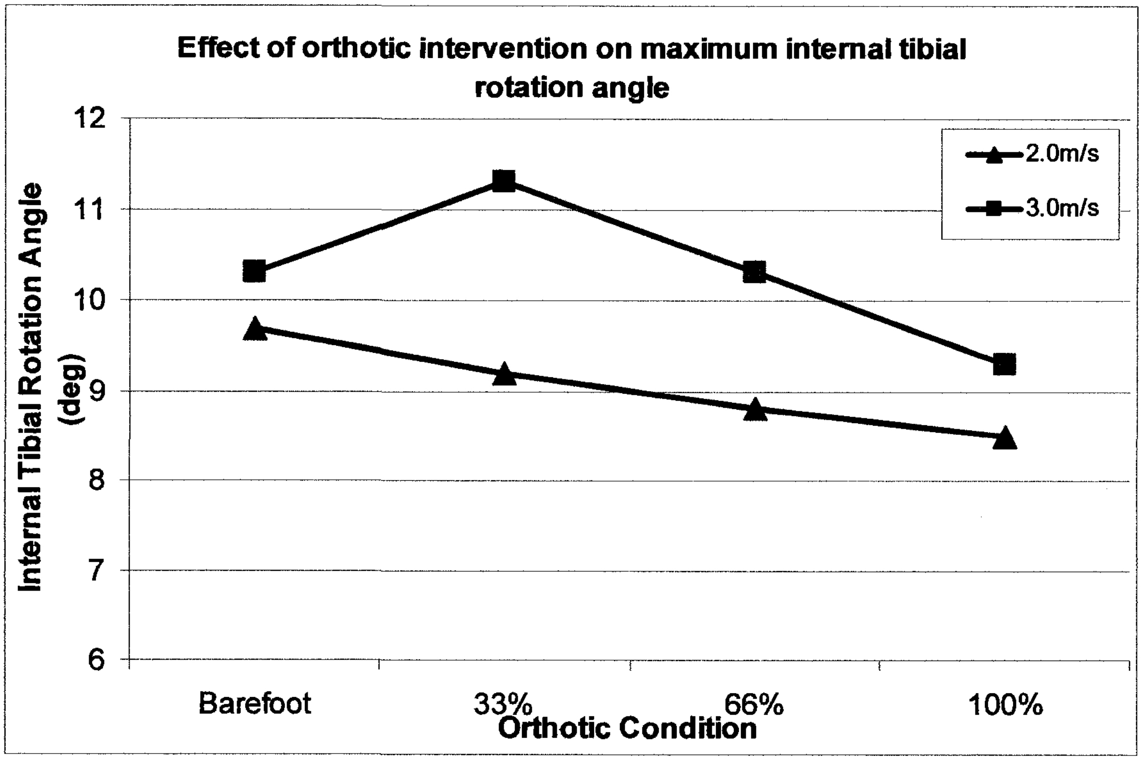

Figure 3.17: The effect of orthotic intervention on maximum internal tibial rotation angle while running at 2.0m/s and 3.0m/s. Mean values are presented; standard deviations are presented on page 136.

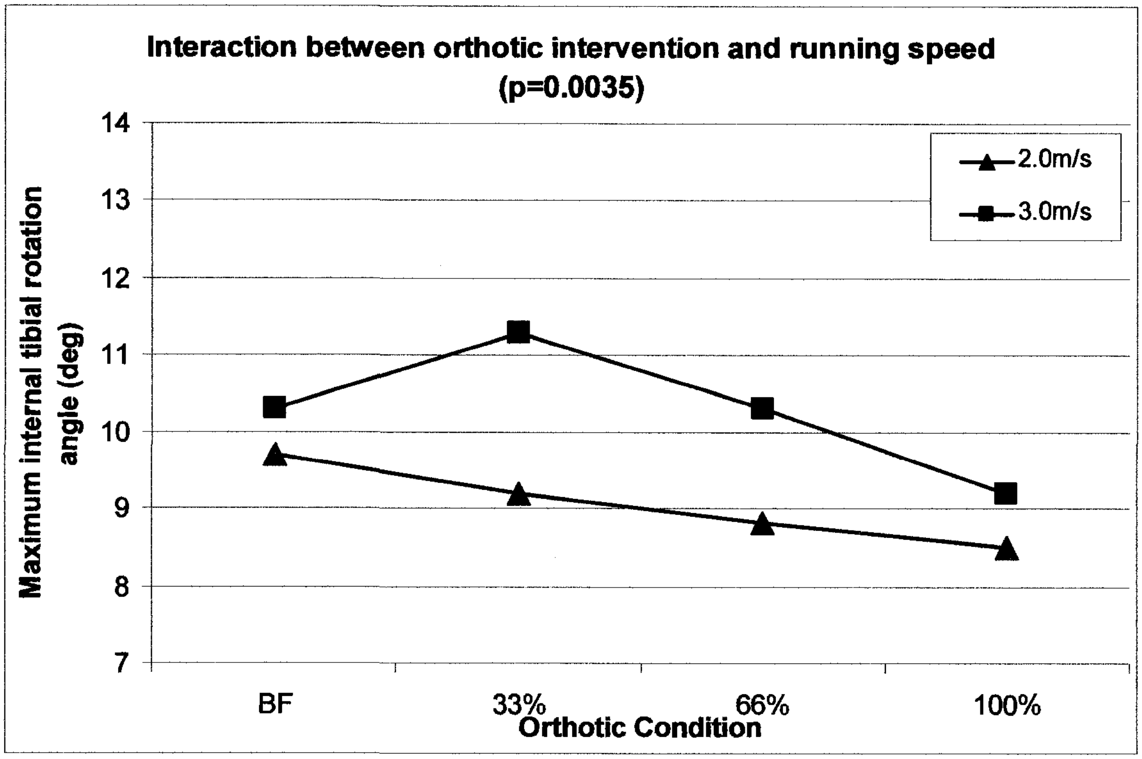

Figure 3.18: The interaction between orthotic intervention and running speed (p=0.0035) on maximum internal tibial rotation angle. Mean values presented; standard deviations are presented on page 136.

3.4.7 Rate of internal tibial rotation

Orthotic intervention did not have a significant effect on the rate of internal tibial rotation ($p=0.2222$). The rates of internal tibial rotation obtained for the BF, 33%, 66% and 100% conditions were 101.7 ± 80.1 deg/sec, 102.9 ± 77.2 deg/sec, 98.2 ± 66.0 deg/sec and 95.8 ± 63.1 deg/sec, respectively (Table 3.6). Running speed had a significant effect on the rate of internal tibial rotation ($p<0.0001$) as 2.0m/s and 3.0m/s demonstrated values of 85.7 ± 60.3 deg/sec and 113.5 ± 79.5 deg/sec, respectively (Table 3.7). The rate of internal tibial rotation was not significantly affected by foot ($p=0.3924$) as the left and right foot produced values of 101.4 ± 64.4 deg/sec and 97.8 ± 79.0 deg/sec, respectively (Table 3.8). Figure 3.19 illustrates the effects of orthotic intervention and running speed on the rate of internal tibial rotation.

No significant interactions were found to occur with the rate of internal tibial rotation.

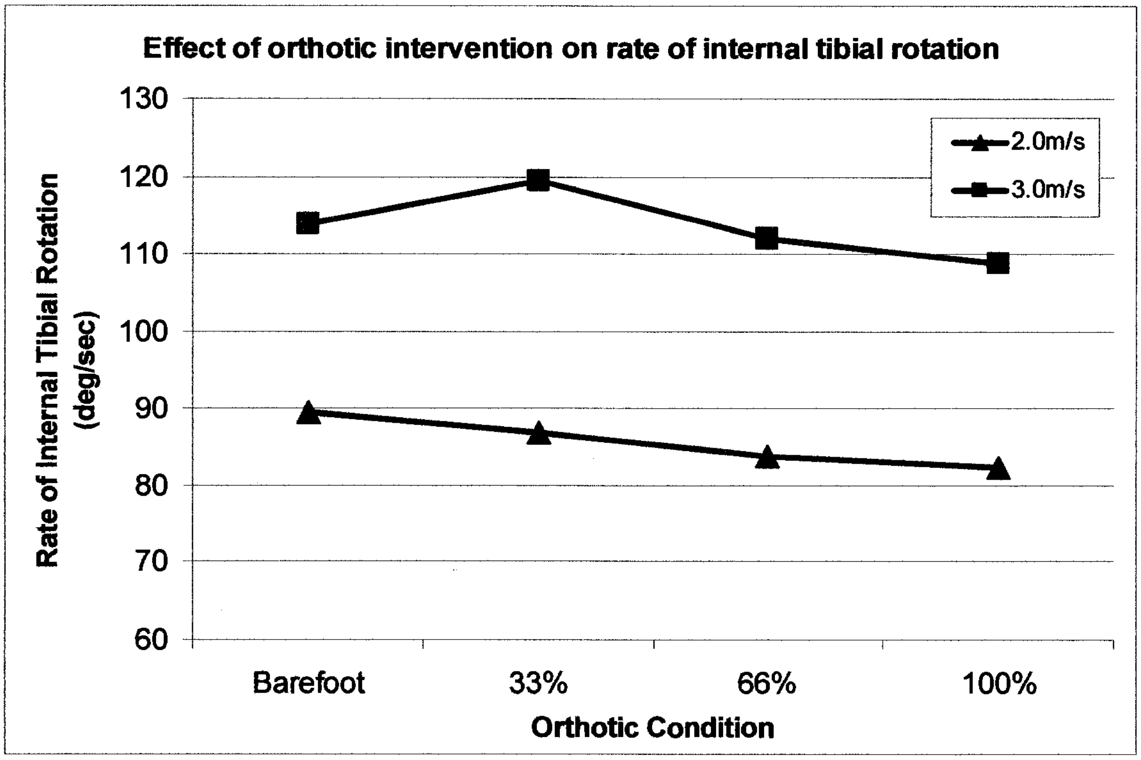

Figure 3.19: The effect of orthotic intervention on the rate of internal tibial rotation angle while running at 2.0m/s and 3.0m/s. Mean values are presented; standard deviations are presented on page 139.

3.4.8 Variability of foot placement during treadmill running

Medioloateral

Participants were found to run on the treadmill with no significant mediolateral discrepancies between foot strikes. Participants consistently landed within 5cm of subsequent foot strikes in this direction.

Anterior-posterior

Participants also demonstrated no significant anterior-posterior discrepancies between foot strikes. Participants consistently landed within 10cm of subsequent foot strikes in this direction.

Table 3.6: The effect of orthotic intervention (BF, 33%, 66%, 100%) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance ($p \leq 0.05$) between BF and 33%
Ω denotes significance ($p \leq 0.05$) between BF and 66%
∞ denotes significance ($p \leq 0.05$) between BF and 100%
§ denotes significance ($p \leq 0.05$) between 33% and 66%
♦ denotes significance ($p \leq 0.05$) between 33% and 100%
δ denotes significance ($p \leq 0.05$) between 66% and 100%

	Barefoot	33%	66%	100%
Max Rearfoot Angle (deg) *Ω∞§♦δ	10.4 (4.0)	10.1 (3.8)	9.4 (3.7)	8.8 (3.6)
Rate of Rearfoot Angle (deg/sec)	115.1 (69.5)	121.8 (76.7)	119.3 (79.1)	114.6 (79.5)
Max Internal Tibial Rotation Angle (deg) Ω∞§♦δ	10 (7.5)	10.2 (8.3)	9.6 (7.6)	8.9 (7.2)
Rate of Internal Tibial Rotation (deg/sec)	101.7 (80.1)	102.9 (77.2)	98.2 (66.0)	95.8 (63.1)

Table 3.7: The effect of running speed (2.0m/s or 3.0m/s) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance ($p \leq 0.05$) between running speed.

	2.0m/s	**3.0m/s**
Max Rearfoot Angle (deg) *	9.2 (3.7)	10.2 (3.9)
Rate of Rearfoot Angle (deg/sec) *	99.2 (61.5)	136.2 (84.8)
Max Internal Tibial Rotation Angle (deg) *	9.1 (7.5)	10.3 (7.8)
Rate of Internal Tibial Rotation (deg/sec) *	85.7 (60.3)	113.5 (79.6)

Table 3.8: The effect of foot (left or right) on the dependent variables. Values represented as mean (standard deviation).
* denotes significance ($p \leq 0.05$) between foot.

	Left Foot	Right Foot
Max Rearfoot Angle (deg)	9.4 (3.1)	9.9 (4.4)
Rate of Rearfoot Angle (deg/sec) *	128.7 (79.7)	106.7 (71.3)
Max Internal Tibial Rotation Angle (deg) *	12.1 (8.2)	7.2 (6.2)
Rate of Internal Tibial Rotation (deg/sec)	101.4 (64.4)	97.8 (79.0)

3.5 Discussion

The objective of this research study was twofold: 1) to determine if individuals with FFF experienced an increase in lower extremity motion during running when compared to individuals with a subtalar neutral foot type and 2) to determine the effects of medial arch supports on lower extremity motion during running among individuals with FFF. We hypothesized that: 1) individuals with FFF will demonstrate an increase in lower extremity motion during barefoot treadmill running when compared to individuals with a subtalar neutral foot type and 2) orthotic intervention will significantly affect lower extremity motion during treadmill running such that barefoot > 33% > 66% > 100% among individuals with FFF.

The results of this study demonstrate no significant differences with respect to maximum rearfoot angle, rate of rearfoot angle and rate of internal tibial rotation between individuals with subtalar neutral and FFF foot types. Surprisingly, maximum internal tibial rotation angle was significantly higher among the subtalar neutral group when compared to the FFF group. Thus, the results do not support the first hypothesis. The results of this study support the second hypothesis as orthotic intervention did have a significant effect on maximum rearfoot angle and maximum internal tibial rotation angle. However, the rate of rearfoot angle and the rate of internal tibial rotation were not significantly affected by orthotic intervention.

This section begins with a discussion of the observed running mechanics between individuals with subtalar neutral and FFF, followed by the orthotic intervention on rearfoot motion and internal tibial rotation. Then, the clinical applications and study limitations are presented followed by recommendations for future research and overall conclusions.

3.5.1 Lower extremity kinematics during running among individuals with subtalar neutral and FFF foot types

No significant differences were observed between the subtalar neutral and FFF groups with respect to age, height or weight. As expected there was a significant difference between groups with respect to static rearfoot angle which was measured to be 4.5 ± 1.1 deg and 11.9 ± 1.3deg for the subtalar neutral and FFF groups, respectively ($p<0.0001$). Therefore, this confirmed that there was a significant difference in static foot structure between the two groups. The increase in static rearfoot angle observed in individuals with FFF is thought to result in excessive motion of the lower extremity during dynamic activities including running. More specifically, excessive pronators have demonstrated an increase in peak rearfoot eversion and tibial internal rotation excursions when compared to normal pronators during running (McClay and Manal, 1997). However, this observation remains controversial within the literature.

The results from the present study suggest that individuals with FFF do not demonstrate an increase in maximum rearfoot angle or rate of rearfoot angle during running when compared to individuals with a subtalar neutral foot type. Thus, our hypothesis was not supported. Similar results have been reported within the literature suggesting that common indicators of flat foot, such as marked static rearfoot angle, are not correlated with maximum rearfoot eversion during walking (Hunt, Fahey and Smith, 2000). These authors concluded that among asymptomatic flat foot individuals, adequate adaptation is possible during walking via muscular compensation. The present study provides evidence for adequate adaptation during running among individuals with asymptomatic FFF as these individuals did not experience excessive lower extremity motion when compared to individuals with subtalar neutral foot types. However, speculation into muscular compensation cannot be confirmed. Studies involving individuals with symptomatic flat foot have also demonstrated similar rearfoot motion during the stance phase of walking when compared to individuals with normal feet (Hunt and Smith, 2004). In addition, investigations involving the relationship between medial arch height and injury suggest that arch height does not influence maximal eversion moment or maximal internal leg rotation during the stance phase of running (Nigg, Cole and Nachbauer, 1993). Thus, a low arch height which is a typical characteristic of FFF may not result in excessive motion of the lower extremity during running.

The results from the present study also indicate that maximum internal tibial rotation was not significantly greater among individuals with FFF when compared to individuals with a subtalar neutral foot type. No differences were observed between the two groups with respect to the rate of internal tibial rotation, however, maximum internal tibial rotation angle was significantly higher among the subtalar neutral group (13.5 ± 8.4 deg/sec vs. 10.0 ± 7.5 deg/sec; $p<0.0001$). Previous research has demonstrated a restraint of motion among individuals with flat feet (Hunt and Smith, 2004). Therefore the running mechanics observed in individuals with FFF may have a protective effect on internal tibial rotation and ultimately knee injuries. This is of particular interest since the knee has been reported to be the most common site to develop a running related injury (Clement, Taunton, Smart and McNicol, 1981). These results are in contrast to other studies which have shown that excessive pronators demonstrate higher internal tibial rotation excursions when compared to normal pronators (McClay and Manal, 1997). These authors reported internal tibial rotation values of 8.9 ± 2.5 deg and 11.1 ± 3.5 deg for the normal and excessive pronators, respectively. The conflicting results may be attributed to the differences in rearfoot angle inclusion criteria into the normal and FFF groups. These authors required rearfoot angles of 8-15 deg and >18 deg to be included in the normal and excessive pronator groups, respectively. The present study required rearfoot angles of 4-6 deg and >10 deg to be included in the subtalar neutral and FFF groups, respectively. Although the differences between groups were similar, it is possible that within the present study a rearfoot angle of >10 deg was not high enough to evoke increases in internal tibial rotation angles.

3.5.2 Orthotic intervention on rearfoot motion among individuals with FFF

The kinematic marker placement and rearfoot motion calculations have been used in previous research to accurately measure rearfoot angle and velocity (Nike, 1989; Kernozek and Ricard, 1990; Perry and Lafortune, 1995; McClay and Manal, 1997; McClay and Manal, 1998; Hetsroni, Finestone, Milgrom, Ben-Sira, Nyska, Mann, Almosnino and Ayalon, 2008). The results of the present study indicate that orthotic intervention had a significant effect on maximum rearfoot angle such that BF > 33% > 66% > 100%. Significance was reached across each orthotic level ($p<0.05$) with values of 10.4 ± 4.0 deg, 10.1 ± 3.8 deg, 9.4 ± 3.7 deg and 8.8 ± 3.6 deg for the conditions of BF, 33%, 66% and 100%, respectively. These findings demonstrate that an increase in support under the medial longitudinal arch results in a decrease in maximum rearfoot angle during running. Similar decreases in maximum rearfoot angle have been found in previous research with the use of foot orthotics ranging from approximately 6 deg to 13.7 deg (Rodgers and Leveau, 1982; Smith, Clarke, Hamill and Santopietro, 1986; Mundermann, Nigg, Humble and Stefanyshyn, 2003; Nester, van der Linden and Bowker, 2003; MacLean, McClay Davis and Hamill, 2006). Mundermann et al. (Mundermann, Nigg, Humble and Stefanyshyn, 2003) demonstrated a significant reduction in maximum rearfoot eversion ($p<0.001$) from 16 deg to 13.7 deg during overground running at 4.0m/s when a medially posted orthotic was inserted into a running sandal. These values are higher than those observed in the present study and may be explained by the faster running speed, greater rearfoot angle inclusion criteria (>13 deg) and footwear within the running conditions associated with the study by Mundermann et al. (Mundermann, Nigg, Humble and Stefanyshyn, 2003). MacLean et al. (MacLean, McClay Davis and Hamill, 2006) demonstrated a similar significant reduction in maximum rearfoot eversion angle during overground running at 3.6m/s while using a custom foot orthotic within a shoe when compared to the shod condition alone ($p=0.025$) although of a lesser magnitude (approximately: shod, 7 deg; orthotic, 6 deg). The smaller magnitude of rearfoot angle may be explained primarily due to the fact that this study included healthy runners with a more optimal static lower extremity alignment when compared to the participants in the present study.

It can be speculated that the observed structural changes, which were recorded during the static rearfoot measurements as a result of orthotic intervention, were consistent throughout dynamic activity. Therefore, orthotics did have a mechanical effect on both static and dynamic maximum rearfoot angles. Currently it is unknown whether this small but statistically significant decrease in maximum rearfoot angle during orthotic intervention would have a significant clinical impact. Arguably, this small reduction in maximum rearfoot angle during a single stance phase may have major clinical implications over time due to the repetitive nature of running.

Orthotic intervention did not have a significant effect on the rate of rearfoot angle ($p=0.2198$). The rates of rearfoot angle obtained for the BF, 33%, 66% and 100% conditions were 115.2 ± 69.6 deg/sec, 121.8 ± 76.7 deg/sec, 119.3 ± 79.1 deg/sec and 114.6 ± 79.5 deg/sec, respectively. Therefore, it appears that orthotics did not have a mechanical effect on the rate of the rearfoot motion during the stance phase of running. Previous studies have also demonstrated no significant decreases in rearfoot velocity during running. Stacoff et al. (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000) completed overground running trials at 2.5-3.0m/s while wearing footwear with the heel counter removed. The differences between subjects were larger than between the orthotic conditions and thus the use of orthotics seem to be highly individualized in terms of rearfoot velocity. The authors reported values ranging from 85 deg/sec to 171 deg/sec between subjects. Although the values from the present study fall within this range, the differences may be due to variations in participant's foot structures or running mechanics. In addition, footwear has been shown to increase foot pronation during running which may also explain the reported increase in rearfoot velocity when compared to the present study (Stacoff, Kaelin, Stuessi and Segesser, 1989). This finding may be of particular interest since the current research suggests that the rate of pronation may contribute to injury more so than the magnitude of pronation (Hreljac, Marshall and Hume, 2000). If this is the case, then orthotics may not provide symptomatic relief through a mechanical mechanism since they did not significantly affect the rate of rearfoot motion.

Conversely, previous research has demonstrated that orthotic intervention does not significantly affect maximum rearfoot angle (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Stackhouse, McClay Davis and Hamill, 2004) and does significantly decrease rearfoot velocity (Mundermann, Nigg, Humble and Stefanyshyn, 2003; MacLean, McClay Davis and Hamill, 2006) which are in contrast to the present study. It is likely that the majority of these differences can be explained by the differences in orthotic construction between research studies. Lack of detail regarding orthotic description within these studies makes it difficult to determine the exact orthotic characteristics that may or may not be responsible for lower extremity changes during running.

3.5.3 Orthotic intervention on internal tibial rotation among individuals with FFF

In order to decrease motion artifact, kinematic markers were placed over areas on the tibia that had little muscle or adipose tissue between the skin and the bone. These concepts for optimal marker placement on the shank have been supported by previous work (Nigg, Khan, Fisher and Stefanyshyn, 1998; Bellchamber and van den Bogert, 2000). In addition, the method used to calculate internal tibial rotation has been used in previous research studies (Perry, 1993). The results of the present study suggest that orthotic intervention significantly decrease maximum internal tibial rotation angle during running among individuals with FFF ($p=0.0019$) as values of 10.0 ± 7.5 deg and 10.2 ± 8.3 deg, 9.6 ± 7.6 deg and 8.9 ± 7.2 deg were associated with BF, 33%, 66% and 100%, respectively. Significant differences were found between each orthotic condition excluding the BF vs. 33% condition. Thus, it appears that orthotics may have a mechanical effect on maximum internal tibial rotation angle. Similar decreases in maximum internal tibial rotation angles with orthotic intervention have been reported in the literature ranging from 3.2 deg to 6.0 deg (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Mundermann, Nigg, Humble and Stefanyshyn, 2003). More specifically, a significant reduction in maximum internal tibial rotation angle was observed from 6deg to 5.5deg (Mundermann, Nigg, Humble and Stefanyshyn, 2003) and from 4.8 deg to 3.2 deg (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000) while wearing foot orthotics. These values, which are lower than those found in the present study, may be explained by their choice of reference frame. These studies measured internal tibial rotation relative to the foot. Since the foot and tibia move together during gait, internal tibial rotation relative to the foot may be less than if it was measured relative to a static structure. The present study measured internal tibial rotation relative to its static position which may explain the increased internal tibial rotation angles found in the present study.

Orthotic intervention did not have a significant effect on the rate of internal tibial rotation ($p=0.2222$). The rates of internal tibial rotation obtained for the BF, 33%, 66% and 100% conditions were 101.7 ± 80.1 deg/sec, 102.9 ± 77.2 deg/sec, 98.2 ± 66.0 deg/sec and 95.8 ± 63.1 deg/sec, respectively. Therefore, it appears that orthotics did not have a mechanical effect on the rate of internal tibial rotation during the stance phase of running. Limited research has examined the effects of orthotic intervention on the rate of internal tibial rotation during running. As such no direct comparisons can be made however, when compared to previous literature the results from the present study seem to be of similar values. In a study involving overground running at 4.0m/s, a maximum tibial rotation velocity of approximately 185 deg/sec was demonstrated (Mundermann, Nigg, Humble and Stefanyshyn, 2003). This increase in velocity when compared to the present study may be attributed to the faster running speed or the fact that the reported velocity may include both internal and external rotations.

In contrast to the results of the present study, previous research has demonstrated that orthotic intervention does not significantly affect maximum internal tibial rotation angle (Nigg, Khan, Fisher and Stefanyshyn, 1998) and does significantly decrease maximum internal tibial rotation velocities (Mundermann, Nigg, Humble and Stefanyshyn, 2003). The conflicting results seen within the literature may be due to the differences in orthotic construction between research studies. In addition, a lack of description regarding orthotic construction makes it difficult to replicate and compare research studies which may also contribute to the variability of the results within the literature.

3.5.4 Clinical applications

The results of the present study indicate that similar lower extremity kinematics exist between individuals with a subtalar neutral foot and FFF during running. In fact, maximum internal tibial rotation angle was significantly higher among the subtalar neutral group when compared to the FFF group. Therefore, clinicians must be careful when using static rearfoot observations to predict dynamic foot function as individuals with increased static rearfoot angles may not experience the expected increase in lower extremity motion during locomotion. In addition, the results of this study provide evidence for the mechanical effect of orthotics as an increase in medial arch support elicited a reduction in rearfoot motion and internal rotation of the tibia. However, no significant decrease in the rates of rearfoot angle and internal tibial rotation was observed with orthotic intervention. Therefore, it is unclear if orthotics function solely through a mechanical mechanism in order to deliver symptomatic pain relief. Further research should investigate other possible mechanisms to explain the positive effects of orthotics. A better understanding of the exact function of orthotics will allow for further modifications by researchers and a more accurate prescription by clinicians in order to improve the quality of life among individuals suffering from abnormal foot mechanics.

3.5.5 Limitations

One possible limitation of this study is that it involved measurement of the running mechanics among individuals with asymptomatic FFF. Since the participants were asymptomatic they may be able to overcome mechanical dysfunction relating to their mal-alignment by means of muscular control. Therefore, even though the participants fit our FFF criteria they may not have shown any dynamic differences when compared to individuals with a more optimal alignment. In addition, it is possible that orthotic intervention may have disturbed the preferred pattern of movement among these individuals resulting in an increase in lower extremity motion. This reasoning may explain why the present study did not find that individuals with FFF exhibited excessive motion of the lower extremity. Further, it may also explain the lack of significance found with the rate of motion at both the rearfoot and tibia.

After dynamic observation it was confirmed that 4 individuals were forefoot strikers, whereas 15 individuals were heel strikers. This difference may limit the results of this study since different running mechanics have been associated with both forefoot and heel strikers (McClay and Manal, 1995). However, current research has demonstrated that orthotic intervention has a similar effect on both forefoot and heel strikers (Stackhouse, McClay Davis and Hamill, 2004). Therefore, it is unlikely that this difference had an effect on the results of the present study.

The construction of the medial arch support used in this study may be a limitation. The maximum height of the arch support was 16mm therefore the 100% condition may not be reflective of the participant's maximum medial arch height if it was greater than 16mm. Further, since the 33% and 66% conditions were constructed based on the measured medial arch height there may not have been much of a difference between the orthotic conditions of 66% and 100% among individuals with a medial arch height greater than 16mm. However, since significance was achieved between these conditions for both maximum rearfoot angle and maximum internal tibial rotation angle it is likely that these two conditions were significantly different. In addition, the arch support used in this study is not similar to a full foot orthotic and therefore cautious interpretation should be employed when attempting to generalize the results of this study to full foot orthotics. As well, participants may have found the medial arch supports more uncomfortable than a full foot orthotic since they did not demonstrate a smooth transition from the heel to the forefoot as is typically seen with a full orthotic. Instead, the arch support was directly under the medial arch and may have caused an abrupt impact during the stance phase of running.

This study attempted to gain a better understanding of the effects of medial arch support during running among individuals with FFF. As a result, footwear was not worn in order to limit confounding results. Due to this decision, this study is limited in its ability to generalize the results since it does not provide an indication of the effects of medial arch supports within a shoe. Therefore, future research should expand on the present study to incorporate the effects of medial arch support within footwear during running.

3.5.6 Recommendations for future work

This study was designed to measure the running mechanics associated with functional flatfoot (FFF) as well as the effects of orthotic intervention during running among this population. Suggestions for future research are listed below:

1. Develop prospective research studies to:
    a. Measure the prevention capabilities of orthotics
    b. Measure the running mechanics associated with FFF and its relation to injury
2. Recruit symptomatic clinical populations for testing
3. Further research into the effectiveness of orthotics through other mechanisms such as:
    a. Sensory feedback from the plantar surface of the foot
    b. Musculature of the lower extremity
4. Examine the long term effects of orthotics on lower extremity kinematics

5. Test the effectiveness of orthotics at running speeds greater than 3.0m/s since this study demonstrated an increase in lower extremity kinematics from 2.0m/s to 3.0m/s.
6. Incorporate a control group into the research design in order to allow for a cause-and-effect interpretation

3.5.7 Conclusions

The controversy surrounding foot orthotic function may be partly attributed to previous methodological designs including footwear and foot type which may have confounded the results. This experiment attempted to control for these possible confounding variables by completing barefoot running trials while controlling for foot type. As a result, this study may provide a better understanding of the basic mechanical effects of foot orthotics.

This experiment demonstrated similar lower extremity running mechanics between individuals with subtalar neutral and FFF foot types. Surprisingly, a significant increase in maximum internal tibial rotation angle was observed among the subtalar neutral group indicating that excessive pronation may have a protective effect on tibial rotation. Therefore, static rearfoot measurement may not be indicative of dynamic foot function of the lower extremity thus, clinicians and researchers should not rely solely on this measurement in order to predict dynamic foot dysfunction. In addition, the present study provided evidence for the effectiveness of orthotic intervention in decreasing maximum rearfoot and maximum internal tibial rotation angles. However, there was a lack of evidence to support the effectiveness of orthotic intervention on the rates of rearfoot angle or internal tibial rotation. This may be of particular importance since lower extremity velocities have been more closely associated with injury than lower extremity angles (Hreljac, Marshall and Hume, 2000). From a mechanical perspective, orthotics may not affect injury prevention since they did not significantly affect the rates of rearfoot angle or internal tibial rotation. However, further prospective studies are required in order to confirm the prevention capabilities of orthotics.

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3.7 Appendices

Appendix 3.1: FFF Criteria

1. STATIC ALIGNMENT	VISUAL INSPECTION	PARTICIPANT'S ALIGNMENT
HIP		
ASIS	Level?	Yes / No
PSIS	Level?	Yes / No
Iliac Crest	Level?	Yes / No
Skin fold on back	Asymmetry present?	Yes / No
Curvature in spine	Excessive?	Yes / No
KNEE		
Genu varum - Knees apart? Genu valgum - Ankles apart?	Stand with knees and ankles together. To participate in the study both ankles and knees must touch.	Knees Touch? Yes / No Ankles Touch? Yes / No
Genu recurvatum - Hyperextension of knee in standing	Stand with knees locked. Is there excessive extension at the knee?	Yes / No
FOOT		
1) Heel – tib/cal angle in weight bearing	Rearfoot angle greater than 10^{o} (Posterior view)	Angle:_______
4) Midfoot Collapse: - Weight bearing - Non-weight bearing	 - Medial longitudinal arch is reduced relative to non-weight bearing - Medial longitudinal arch is present	 Yes / No Yes / No

3) Forefoot – weight bearing	Abduction?	Yes / No
2. DYNAMIC ALIGNMENT	VISUAL INSPECTION	PARTICIPANT'S ALIGNMENT
Heel Toe Running		Yes / No
Symmetry between left and right	Running	Yes / No
Heel Strike	Lateral (supinated)? OR Medial (pronated)?	Yes / No Yes / No
Midstance	Excessive pronation? OR Prolonged pronation?	Yes / No Yes / No
Toe-off	Re-supination? OR Continued pronation?	Yes / No Yes / No
3. ORTHOTIC INTERVENTION	Confirming structural changes due to orthotic intervention	

Testing Condition	Calcaneal Eversion (degrees)	
	R	L
Barefoot		
33% Arch Height		
66% Arch Height		
100% Arch Height		

Perceived Comfort of Orthotic	
Most comfortable condition	
Least comfortable condition	

Appendix 3.2: Screening Questionnaire

VOLUNTEER EXCLUSION CRITERIA **Date: (MM/DD/YYYY): ___, ___, ______**

Name: ____

Address: ____

____

City, Province: _______________, _____ Postal Code _______________

Tel #: (____)-___________ Best time to call: ___________

Age: ____ yrs. Height: ______ cm Weight: _______ kg

Gender: M __ F __

Do you have any conditions that limit the use of your legs? Yes / No

If yes, how much does the condition interfere with your activities?

little or none	moderate	a great deal
€	€	€

Describe: ____

____

____

Do you have or have you ever had: **Yes / No**

		Yes / No
a)	paralysis	___
b)	epilepsy	___
c)	cerebral palsy	___
d)	multiple sclerosis	___
e)	Parkinson's disease	___
f)	stroke	___
g)	any other neurological disorder	______________________
h)	diabetes	___

Have you ever had frostbite in the lower extremities? **Yes / No**

How much do the conditions that you indicated with a 'yes' below interfere with your activities?

Do you have or have you ever had :

		Yes / No	little or none	moderate	a great deal
a)	problems with your heart or lungs	___	€	€	€
b)	high blood pressure	___	€	€	€
d)	blood circulation problems (generally)	___	€	€	€
	(specifically lower extremities)	___	€	€	€
d)	cancer	___	€	€	€
e)	arthritis	___	€	€	€
f)	rheumatism	___	€	€	€
g)	back problems	___	€	€	€
h)	a joint disorder	___	€	€	€
i)	a muscle disorder	___	€	€	€
j)	a bone disorder	___	€	€	€
k)	spina bifida	___	€	€	€

How much do the conditions that you indicated with a 'yes' below interfere with your activities?

Do you have or have you ever had these <u>foot</u> problems:

		Yes / No	little or none	moderate	a great deal
a)	bunions (hallux valgus)	___	€	€	€
b)	hammer toes	___	€	€	€
c)	calluses	___	€	€	€
d)	ulcerations	___	€	€	€
e)	plantar fasciitis	___	€	€	€
f)	any other foot problems (diagnosed or not)				
	______________________	___	€	€	€
	______________________	___	€	€	€
	______________________	___	€	€	€

Have you ever severely injured or had surgery on your (specify)

		Yes / No	little or none	moderate	a great deal
a)	ankle	___	€	€	€
b)	knee	___	€	€	€

c)	hip	___	€	€	€
Have you ever broken any bones?		___	€	€	€

Which ones? : ______________________________

Have you had any recent (specify)

a)	illnesses	___	€	€	€
b)	injuries	___	€	€	€
c)	operations	___	€	€	€

Do you have difficulties performing any daily activities?	___	€	€	€

Which activities?: ______________________________

Are you currently taking any medications (prescription or over-the-counter), or other drugs?

Medication	Ailment	Frequency of use
____________	____________	____________
____________	____________	____________
____________	____________	____________

<u>**Orthotic Use**</u>

Do you have foot orthotics? **YES/NO**

	less than 1day/wk	2-3 days/wk	more than 5days/wk
If answered YES: **How often do you wear your othotics?**	€	€	€

	less than 2hrs	2-5hrs	more than 5hrs
How long do you wear your orthotics for at one time?	€	€	€

What types of activities require orthotic use? Describe below.

____

____

____

____

____

Appendix 3.3: Medial Arch Support Construction

1. A left and right foot casting was completed during Session 2 by a Certified Pedorthist for each participant. From this cast, a foot mold was created.
2. A 29 mm ruler was laid flat on top of the foot mold such that it rested on the first metatarsal head on one end and the medial calcaneus on the other.
3. Under the area of the ruler, the point at which the arch appeared highest was estimated. A small pen mark was placed at this point.

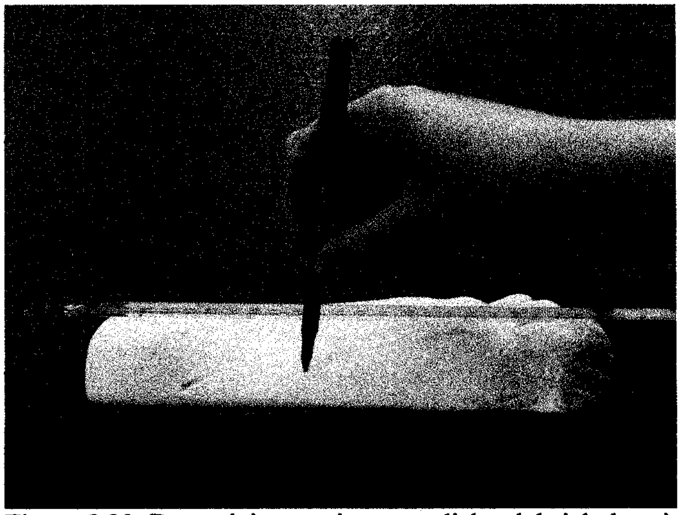

Figure 3.20: Determining maximum medial arch height location.

4. The mold was then positioned such that the sole of the foot was in contact with the table and the medial edge of the foot was in line with the edge of the table.
5. An electronic caliper with digital display (Mastercraft) was positioned in line with the mark made previously on the foot mold indicating the maximum arch height. The distance from the table to this mark on the arch was measured and recorded. The electronic caliper was placed at the same distance from the edge of the table (1 cm) for each subject to ensure consistency.

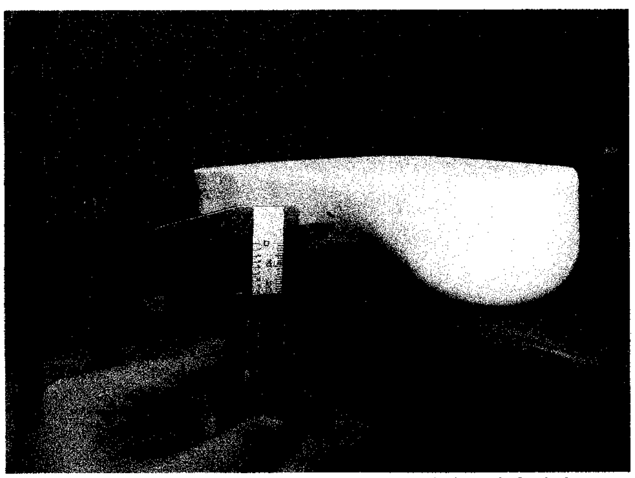

Figure 3.21: Measuring maximum medial arch height.

6. The maximum arch height was rounded down to the nearest whole millimeter.
7. One-third and two-thirds of the maximum value were calculated using excel. These values were rounded to the nearest whole millimeter.
8. The distance between the table and the maximum height of the arch cookie was measured using the caliper. The arch cookies were all 16 mm in height thus, if a participant's maximum arch height value exceeded 16 mm, the full arch cookie was used as the maximum value. The arch cookies for the 33% and 66% arch height conditions were still calculated from the maximum measured arch height.
9. The distance on the calipers was set to the difference between the maximum height of the arch cookie (16 mm) and the desired arch cookie. For example, if you were trying to create an arch cookie with a height of 9 mm, the calipers would be set at 7 mm.
10. The edge of the caliper was used to score a line across the arch cookie. This line was then drawn on in pen.

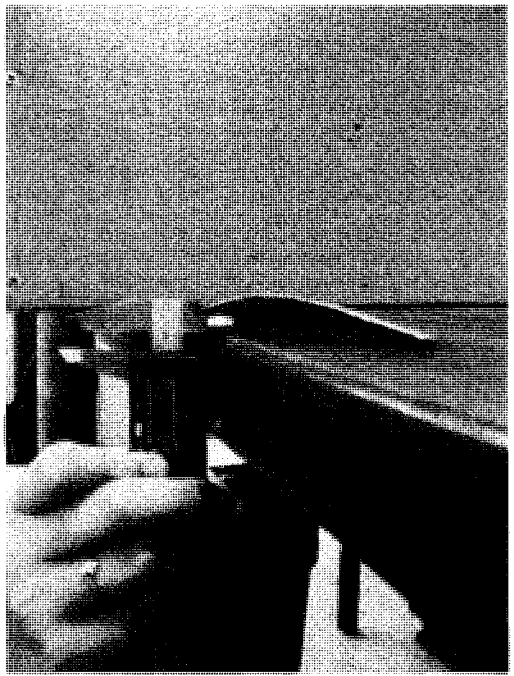

Figure 3.22: Creating the appropriate arch height using the electronic digital caliper.

11. The ends of the line were indicated on the top of the arch cookie. At the midpoint of these two markings, a compass was centered and a half-circle was drawn connecting the two initial markings made on the top of the arch cookie (as a continuation of the line etched on by the caliper).
12. An electric sander was then used to sand the bottom of the arch cookie down to the level of the scored line.

CHAPTER 4: GRAND DISCUSSION

4.1 Introduction

The main purpose of this thesis was to investigate the effects of orthotic intervention during running among individuals with functional flatfoot (FFF). Since orthotics are typically prescribed to this clinical population for rehabilitative purposes it is critical to understand how orthotics affect these individuals. Substantial clinical and anecdotal evidence exists for the effectiveness of orthotics in providing symptomatic pain relief however the underlying mechanism by which orthotics function continues to remain controversial in the literature. This study was designed to investigate the mechanical effects of medial arch supports on lower extremity kinematics during running in order to gain a better understanding into the mechanism behind orthotic function. Chapter 3 outlines the justification and methodology behind this research project as well as the results and discussion.

In order to use a treadmill during the main thesis project an initial study was conducted to determine the accuracy of treadmills in representing the lower extremity kinematics associated with overground running among individuals with a subtalar neutral foot. Chapter 2 describes the justification and methodology behind this research study as well as the results and discussion. In addition, the results from this study were compared to the results of the main thesis project to determine if individuals with FFF experienced different lower extremity kinematics during barefoot treadmill running when compared to individuals with a subtalar neutral foot type. The findings are outlined in Chapter 3. The following sections will summarize the major findings from the previous chapters.

4.2 Treadmills accurately represent overground running

This research study examined the barefoot running mechanics associated with treadmill and overground running among individuals with a subtalar neutral foot type. It was hypothesized that treadmills accurately represent the lower extremity kinematics associated with overground running.

The results, as presented in Chapter 2, demonstrate similar kinematic values between treadmill and overground running with respect to the rate of rearfoot angle, maximum internal tibial rotation angle and the rate of internal tibial rotation. A number of other studies have demonstrated similar lower extremity kinematics between treadmill and overground running (Ingen Schenau, 1980; Schache, Blanch, Rath, Wrigley, Starr and Bennell, 2001; Lavcanska, Taylor and Schache, 2005) and walking (Matsas, Taylor and McBurney, 2000). However, maximum rearfoot angle was significantly higher during treadmill running (9.7 ± 3.3 deg) when compared to overground running (8.8 ± 2.9 deg), although this difference was less than 1 deg. Similar rearfoot angles ranging from 8.2 to 11.2 deg during running have been recorded within the literature (McClay and Manal, 1997; McClay and Manal, 1998; Nigg, Khan, Fisher and Stefanyshyn, 1998; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). From this range it appears that rearfoot angles obtained during overground running are associated with the values at the lower end of the range (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000). Thus, the previous literature suggests that treadmill running may produce increased rearfoot angles however these studies did not compare this variable between running surfaces. Therefore, these differences may be a result of differences within running speed or participant selection between studies. It is currently unknown whether the small statistically significant increase in rearfoot angle observed during treadmill running in the present study would be of clinical significance. However, if clinical or experimental decisions are dependent on small changes in maximum rearfoot angle then cautious interpretation should be employed when using treadmills. Otherwise, the results from this study indicate that treadmills accurately represent the lower extremity kinematics associated with overground barefoot running at speeds of 2.0m/s and 3.0m/s.

4.3 Individuals with subtalar neutral and FFF foot types experience similar lower extremity kinematics during running

This experiment investigated the running mechanics associated with individuals with different foot types. More specifically, lower extremity kinematics were examined between individuals with subtalar neutral and FFF foot types while running on a treadmill. It was hypothesized that individuals with FFF would exhibit an increase in lower extremity motion during running when compared to individuals with a subtalar neutral foot type.

The obtained results are described in Chapter 3 and indicate similar running mechanics between the two groups. In particular, the FFF group did not experience a significant increase in maximum rearfoot angle, rate of rearfoot angle, maximum internal tibial rotation angle or rate of internal tibial rotation. Thus, this study did not provide support for our hypothesis. Similar results have been demonstrated within the literature suggesting that no significant differences exist between flat and normal feet during walking (Hunt and Smith, 2004) or running (Nigg, Khan, Fisher and Stefanyshyn, 1998). It is interesting to note that the subtalar neutral group demonstrated an increase in maximum internal tibial rotation angle during running when compared to the FFF group. It has been speculated that individuals with FFF may have more of a restraint of motion as opposed to excessive motion (Hunt and Smith, 2004) during locomotion. Therefore, the condition of FFF may have more of a protective effect on maximum internal tibial rotation angle during running which may result in a lower risk of developing a knee injury. It may be speculated that individuals with asymptomatic flatfoot may have greater strength of their quadriceps and hamstring muscle groups which may result in less tibial rotation when these muscles co-contract. This may be of particular interest to runners as the knee has been found to be the most common site by which to develop a running related injury (Clement, Taunton, Smart and McNicol, 1981).

In addition, it has been suggested that common indicators of flat foot, such as static rearfoot angle, may not be associated with an increase in maximum rearfoot eversion during walking (Hunt, Fahey and Smith, 2000). Therefore, individuals with asymptomatic FFF may adequately adapt during dynamic movement in order to overcome their static foot dysfunction. This adaptation may be occurring through lower extremity muscular compensation. More specifically, the muscles involved with foot supination may be stronger among individuals with asymptomatic flatfoot in order to correct for the excessive foot pronation that is typically seen among this symptomatic population. This study provides evidence for adequate adaptation among individuals with asymptomatic FFF during running since no significant increase in lower extremity kinematic variables were demonstrated. Further studies involving individuals with symptomatic FFF may be required to better understand the running mechanics associated with individuals who may benefit from orthotic intervention.

4.4 Orthotic intervention decreases the magnitude of rearfoot and internal tibial rotation angles during running

The main objective of this thesis project was to determine the effects of orthotic intervention on the lower extremity kinematics during running among individuals with FFF. Chapter 3 further describes the four orthotic conditions that were tested during running as well as the results. Briefly, the orthotic conditions included barefoot (BF) and 33%, 66% and 100% of the participant's maximum medial arch height. It was hypothesized that orthotic intervention would demonstrate a decrease in rearfoot and internal tibial rotation motions during running among individuals with FFF such that 100% < 66% < 33% < BF.

The results demonstrated a significant and systematic decrease in maximum rearfoot angle and maximum internal tibial rotation angle during running between each orthotic condition. More specifically, 100% < 66% < 33% < BF with significance reached between each orthotic condition excluding the BF vs. 33% condition during maximum internal tibial rotation angle only. Previous research has demonstrated similar decreases in rearfoot motion (Mundermann, Nigg, Humble and Stefanyshyn, 2003; Nester, van der Linden and Bowker, 2003; MacLean, McClay Davis and Hamill, 2006) and internal tibial rotation (Nawoczenski, Cook and Saltzman, 1995; Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000; Mundermann, Nigg, Humble and Stefanyshyn, 2003) with the use of foot orthotics. Of debate is whether these small but statistically significant decreases in lower extremity angles between orthotic conditions would have clinical implications. Certainly, if one was to consider the repetitive nature of running, the demonstrated results may be of great clinical importance.

In addition, the obtained results demonstrated no significant differences in the rates of rearfoot angle or internal tibial rotation with orthotic intervention. Similar results have been reported within the literature for maximum rearfoot velocity (Stacoff, Reinschmidt, Nigg, van den Bogert, Lundberg, Denoth and Stussi, 2000) however, limited research has investigated the effects of maximum internal tibial rotation velocity. These non-significant findings may be a result of the participant criteria as this study recruited individuals with asymptomatic FFF. Based on the fact that these individuals present with FFF however are asymptomatic, this indicates that they may be able to adequately overcome their increased static rearfoot angle, resulting in a more optimal dynamic movement. As such, orthotic intervention may not have had an effect on these individuals because they demonstrated optimal dynamic alignment even in the barefoot condition. In addition, it has been suggested that orthotics may cause a disruption to the preferred path of movement, resulting in an increase in variability or total motion of the lower extremity (Stefanyshyn and Hettinga, 2006). Among individuals with asymptomatic FFF, it appears that this may have been the case for the rates of rearfoot angle and internal tibial rotation. This may be of clinical importance in terms of injury prevention as it has been demonstrated that the rate of pronation may contribute to injury more so than the angle of pronation (Hreljac, Marshall and Hume, 2000). Thus, orthotics may not have a large impact on injury prevention. Future research on individuals with symptomatic FFF is required in order to understand the effectiveness of orthotic intervention among this clinical population. In addition, the effects of orthotic intervention on subtalar joint displacement should be examined since changing the axis of this joint may affect the transfer of foot pronation to internal tibial rotation.

4.5 Conclusions

The results of this thesis provide evidence for the effectiveness of orthotics in decreasing maximum lower extremity angles during running among individuals with FFF. However, orthotic intervention did not seem to have a significant effect on the lower extremity rates during running among this population which may have further implications on injury prevention. In addition, individuals with FFF did not demonstrate excessive motion of the lower extremity when compared to individuals with a subtalar neutral foot type. Conversely, maximum internal tibial rotation angle was significantly higher among the subtalar neutral group. This indicates that perhaps the condition of FFF may have some protective benefits in terms of injury however further research is needed in order to confirm this finding. In terms of running surface, treadmills were found to accurately represent the lower extremity kinematics associated with overground running. Although a significant increase in maximum rearfoot angle was observed with treadmill running, it is unclear whether this small difference would be of clinical significance. However, careful interpretation of treadmill results should be employed if clinical or experimental decisions are dependent on small changes in rearfoot motion.

4.6 References

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