Effects of Orthotic Wear on the Kinetic, Kinematic and Electromyographic Characteristics of Walking and Running

by
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Abstract

Although custom-made foot orthotics are commonly prescribed to relieve lower limb injuries, few studies have documented their effects on the biomechanics of locomotion. The objective of this project was to quantify the effect of orthotic wear on kinematic, kinetic and electromyographic characteristics of the legs during walking and running. Fourteen subjects with custom-made foot orthotics were asked to run and walk over a 10-m walkway. Kinematic, kinetic and electromyographical parameters were recorded during all trials. One-way repeated measures ANOVA and paired students t-tests were used to evaluate the effect of orthotic wear as well as foot type (flat, normal). With orthotic wear, the activity of the soleus muscle was decreased for both groups of subjects with and without flat feet during running. The effect of orthotic wear on all other parameters was not significant. More in-depth studies are needed to generalize these results on the overall population of orthotic wearers.

Résumé

Bien que les orthèses plantaires soient régulièrement prescrites afin de diminuer les blessures aux jambes, peu d'études ont documenté leur effet sur la biomécanique de la locomotion. L'objectif de ce projet était de quantifier l'effet du port d'orthèses plantaires sur les paramètres cinématiques, cinétiques et électromyographiques des jambes durant la marche et la course. Quatorze sujets portant des orthèses devaient courir ou marcher sur une distance de 10 m. La cinématique, la cinétique et l'électromyographie ont été enregistrées lors des essais. Des mesures répétées d'analyse de variance et des tests T appariés ont été utilisés pour évaluer l'effet du port d'orthèses et du type de pied (plat, normal). Les résultats démontrent que l'activité du muscle soleus diminue avec le port d'orthèses durant la course. L'effet du port d'orthèses sur les autres paramètres est non-significatif. Des études approfondies sont nécessaires afin de généraliser les résultats sur la population de gens portant des orthèses.

Glossary

Biceps femoris: Muscle of the leg whose main action is knee flexion and internal rotation of the leg. It is part of the hamstring muscle group.

Dorsiflexion: Flexion at the ankle joint where the sole of the foot is facing upwards.

Electromyography: The detection of the electrical activity of a specific muscles' action potential.

Eversion: A movement at the ankle joint where the sole of the foot is rotated laterally.

Extension: Increase in the joint angle.

Flexion: Decrease in the joint angle.

Foot Orthosis: An orthotic device that is used for correcting biomechanical problems that may exist at the foot that can potentially lead to an injury.

Gait: A way to describe a method of locomotion or walking style.

Ground reaction force: Equal and opposite force applied by the foot to the ground.

Inversion: A movement at the ankle joint where the sole of the foot is rotated medially.

Kinematics: The study of movements.

Kinetics: The study of forces.

Lateral gastrocnemius: Muscle of the lower leg that helps to plantar flex the ankle. It is part of the triceps surae group.

Medial gastrocnemius: Muscle of the lower leg that helps to plantar flex the ankle. It is part of the triceps surae group.

Peroneus longus: Muscle of the lower leg whose main action is eversion and plantarflexion of the foot.

Pes planus: A term described as flat feet, which is due to a laxity of the medial longitudinal arch in the foot.

Plantarflexion: A movement at the ankle joint where the sole of the foot is facing downwards.

Pronation: A combination of eversion and abduction of the foot.

Run: A method of locomotion which includes a stance, swing and two float phases where during the float phase the both legs are completely airborne.

Soleus: Main ankle plantarflexor.

Stance: The stance phase is a subdivision of the gait cycle, which occupies approximately 65% of the cycle and consists of two periods of double limb support.

Subtalar joint (talo-crural joint): This joint is located between the talus and the calcaneus. Eversion and inversion movements occur at this ankle joint.

Swing: The swing phase occupies approximately 35% of the gait cycle. It is the action during the gait cycle when the swinging contralateral foot is airborne and prepares to make initial heel strike contact with the ground.

Tibial inclination: The degree of angular displacement of the tibial segment.

Tibialis anterior: Muscle of the lower leg whose main action involves dorsiflexion and invertion of the foot.

Vastus lateralis: Muscle of the quadriceps muscle group whose main action is to extend the knee and external rotation of the leg.

Vastus medialis: Muscle of the quadriceps muscle group whose main action is to extend the knee and internal rotation of the leg.

Walk: A method of locomotion involving a stance and a swing phase where at least on leg maintains contact with the ground throughout this movement.

Chapter 1- Introduction

1. Introduction

Walking is a complex everyday task that many people often take for granted. Walking can be defined as a method of locomotion involving both legs destined to propel the body in a forward translation (Whittle, 2003). Moving from one place to another often involves some form of locomotion either by walking or running. The purpose of locomotion is to ensure a safe and efficient translation of the body over different surfaces at different inclines (Winter, 1987).

Different types of footwear are needed in walking or running whether these are performed in the context of sports or in the realm of everyday activities. Proper shoe selection and fit are essential to all aspects of daily living such as walking, running or climbing. Irregularities found at the ankle, knee or hip joint can result in inefficient walking or running patterns. In such cases the use of orthotics or foot insoles are often prescribed. Whether the person is known to have high or low foot arches or patellofemoral syndrome for example, orthotic intervention is usually offered to correct the alignment of the lower leg for a more efficient way to walk or run.

An orthotic device is used for correcting biomechanical problems that may exist at the foot that can potentially lead to an injury. Orthotics can be made out of plastic, thermoplastic, rubber, sorbethane or leather and they are placed in the shoe as a replacement for the existing insole (Arnheim and Prentice, 2002). In general, foot orthotics (or orthoses) are prescribed to serve as an ergogenic aid, to increase the overall comfort, improve performance and to provide support in the rehabilitation process. However, health care professionals typically perform their assessment while the patient is standing, and little is known about how the orthotics act during dynamic tasks such as walking. Moreover, foot orthotics are commonly prescribed by health care professionals to runners to prevent or help alleviate the symptoms of running related injuries. However, little is known about the effects of foot orthotic intervention on the actual biomechanical characteristics of the lower limbs during walking and running, which constitutes the scope of this thesis. In turn, this kind of knowledge could contribute to a better understanding of the role of orthotic wear during gait and could also provide quantitative evidence to support recommendations for orthotic wear.

Chapter 2- Review of Literature

2. General walking and running biomechanical characteristics

There are many biomechanical differences between walking and running. Walking involves a cyclical motion that is composed of both stance and swing phase components. For each leg, the stance phase occupies approximately 65% of the stride (Nordin and Frankel, 2001). The gait cycle can be divided into six events or periods. 1) Initial contact or heel strike, which is visually seen the instant the heel makes contact with the ground. 2) The loading response is the interval where the sole of the floor meets the ground. 3) Midstance occurs when the tibia rotates over the leading foot. 4) Terminal stance is the period where the weight of the body is transferred from the hind and midfoot towards the forefoot. 5) Pre-swing occurs at the same time as terminal double limb support. The end of pre-swing happens at the same time as toe off. 6) The swing phase occupies the remaining 35% of the gait cycle. Its sub phases, which include initial swing, mid-swing and terminal swing help to illustrate the action that the swinging foot is going through as it is in the air and prepares to make initial contact with the ground for heel strike of the same limb.

Running, on the other hand, consists of four phases: the stance, float, swing and a second float phase. The stance phase in running decreases to 40% of the cycle. The float phase occurs when both legs are airborne and the body is completely off the ground. The swing phase occurs when there is contralateral heel contact that quickly goes into the swing phase. At the swing phase the body prepares to be propelled into the second float phase, where once again the body is completely airborne. Even though both running and walking may appear to involve the same movement; there are noticeable differences between the two. Mostly the appearance of the flight phases in running whereas walking consists of an extended stance phase.

2.1. Walking biomechanics

In walking, the muscles of the stance ankle are used to control the position of the leg as it moves forward by adjusting medio-lateral or anterioposterior acceleration of the body's center of mass (Winter, 1987). The swing leg helps to prevent the body from falling forward at each step of walking (Winter, 1987). During locomotion, the body is also trying to maintain its balance by moving from side to side. As the person's body weight shifts from left to right, the base of support is distributed in the medio-lateral direction while the body is moving forward. Upon heel contact, the pelvis, femur and tibia rotate internally to support the body. At initial contact through midstance, the subtalar joint everts while the foot pronates and the forefoot becomes flexible enough to absorb shock. These movements of the foot cause internal rotation of the higher leg segments during this phase. As the tibia internally rotates, the subtalar joint everts. This is a compensatory movement of the lower extremity in early stance to coincide with the ankle joint axis movement (Nordin and Frankel, 2001). As the body progresses towards the end of the stance to the swing phase, the tibia rotates externally. The foot now becomes a rigid structure that is ready to propel the body forward.

The biomechanics of walking appear to be a complex task that requires control of the lower limbs. During the stance phase, the quadriceps muscle group helps to counteract flexion of the knee due to the mass of the upper body as it collapses the knee joint. The quadriceps muscle group is activated during terminal swing and produces an eccentric contraction during weight acceptance as the knee rotates from a full-extended position at initial contact to its peak support phase flexion of approximately 20 degrees during the loading response (Nordin and Frankel, 2001).

In addition, the muscles around the knee joint are mostly activated during initial contact where the quadriceps muscle group extends the knee joint. The hamstring muscles are activated in late mid-swing. Their main function at the knee is to control the knee joint as it goes into extension. The short head of the biceps femoris is activated earlier than the other hamstring muscles in early mid-swing to assist in flexing the knee as it clears the floor. Analyzing the extensor and flexor knee muscles in the stance phase can be useful in determining how the quadriceps and hamstring muscles respond to orthotic use. Discovering the differences between orthotic and non-orthotic conditions in walking is helpful in finding what changes occur at the lower leg on a whole. By determining these changes enables researchers and health care professionals to quantify results for orthotic prescription.

2.2. Running biomechanics

Running is a widely used skill in physical activity. Whether it is done on its own or integrated in a sport, running is a common but complex skill. The mechanics of running consist of support phases and airborne phases. Upon initial contact, the foot hits the ground with the lateral aspect of the shoe (McKenzie et al., 1985). Due to the fact that there is some inter-subject variability in running patterns, initial contact with the midfoot and forefoot region may also be possible. In general, at initial contact, the ankle pronates the foot. This action assists to absorb the initial contact forces, which allow the foot to adapt to the running surface (i.e. incline, smooth or rigid surfaces). Rapid flexion of the hip, knee and ankle dorsiflexion also occur upon ground contact to help to absorb the impact of the foot striking the ground. Ankle dorsiflexion and knee flexion last for approximately 55% of the support phase and act to absorb the ground reaction forces. During the swing phase, the body's center of gravity moves in front of the stance leg as the swing leg progresses. The femur and the tibia are externally rotated which brings the foot back to supination. Towards the end of the swing phase (approximately 80% of the running cycle), the hip and knee are extended and the ankle plantarflexes until toe-off is attained. Overall, the joints involved in the running cycle are useful in absorbing shock that is placed on the lower limb.

2.3. The role of orthotics in walking and running

As seen in the literature, several reasons typically motivate the use of orthotics in walking or running-related activities. One of the proposed roles of an orthotic device is to help individuals produce a smooth and efficient translation of the body's center of mass during locomotion. Injury may lead to the decrease of one's ability to freely move around on one's own. Another role of foot orthotics is to help restore dynamic stability and reduce excessive pronation of the subtalar joint during stance (McCulloch et al., 1993). In addition, they are believed to help in correcting abnormal alignment of the skeleton to produce a more efficient way of locomotion. Orthotics are used mainly in running-related tasks, however they can be used in a rehabilitative manner for specific populations such as individuals with osteoarthritis (Maly et al., 2002) and rheumatoid arthritis (Woodbrun et al., 2003). Another hypothesis posed in the literature is that foot orthotics serve to improve sensory feedback by increasing the area of contact between the foot sole and the shoe (Nigg et al., 1999). In addition to preventing lower extremity injuries, orthotics are thought to help reduce the range of pronation, reduce internal tibial rotation and provide better alignment of the skeletal system. They have also been previously used in managing the biomechanical effects of pathologies in those who suffer from osteoarthritis of the knee or patellofemoral syndrome. Overuse injuries are most common in sporting activities, mainly running (Razeghi and Batt, 2000). These injuries may include medial tibial stress syndrome, achilles tendonitis, and plantar fascitis. Injuries that occur due to excessive supination include iliotibial band friction syndrome, trochanteric bursitis, anterolateral tibial stress syndrome, achilles tendonitis, plantar fascitis and peroneal tendonitis (Benkeboom et al., 2000). According to Bordelon (1989), orthotics reduces impact pressures by providing support adjacent to the area of pressure. Indeed, they are designed to act to reduce shear forces under the foot by using shock-absorbing materials, reduce ground reaction forces, and alter muscle activation patterns and knee joint movements. Despite all these reports, many of these characteristics have yet to be quantified through further research. By recording and analyzing how the body shifts from left to right in the frontal plane during running or walking, we can better evaluate the effectiveness of orthotics in reducing the mediolateral leg motion and foot forces.

2.4. Orthotics and walking

2.4.1 Kinetics

Force plates are used to measure the ground reaction force (GRF) beneath the foot during the stance phase. The force plate outputs the total force applied by the foot to the ground (Whittle, 2003). Perry and Lafortune (1995) studied the effects of pronation on loading impact. The purpose of this study was to quantify any changes in impact loading due to orthotic wear during running and walking. There were three conditions administered: without orthotics, a varus wedged or valgus wedged orthotic. Authors found that in walking, there was a minimal increase in pronation compared to running. Overall during walking, they found no differences in walking GRF between the three tested conditions. Nester et al. (2003) found that the lateral force increased with the medially wedged orthotic and decreased with the laterally wedged orthotic. Moreover, with the medial wedged orthotic, the authors observed a decrease in pronation. With the lateral wedged orthotic, there was an increase in pronation. Compared to walking without orthotics, medial or lateral wedged orthotics showed some changes in the rearfoot complex pronation, which is associated with internal rotation of the leg relative to the foot. The ground reaction force data obtained from this study showed that one of the shock absorption phases (the contact phase) showed reduced pronation while using the medially wedged orthotic. Since most of the shock absorption properties occurred in the medial/lateral force directions, the authors mention that the vertical and anterior/posterior were less likely to be affected with orthotic use since orthotics were more efficient in reducing pronation/ supination and also internal/ external rotation of the leg.

Overall, these studies suggest that evaluating the ground reaction forces during walking is useful in better understanding the effects of orthotic wear on the impact sustained by the lower leg structures, particularly in the medio-lateral direction. Overall, not many studies have quantified the medio-lateral forces and the effects of orthotic wear during walking. We hypothesize that the effects of orthotic wear will be most clearly highlighted when measured using all the parameters previously mentioned together.

2.4.2. Kinematics

Another way to analyze the effects of orthotic use and evaluating its effect on walking could be to quantify the characteristics of lower limb motion. Two studies (Eng and Pierrynowski, 1994; Gross and Foxworth, 2003) found that with orthotics, there was a decrease in the knee joint motion in the frontal plane during initial contact in walking. Eng and Pierrynowski studied ten female adolescents who were diagnosed with patellofemoral pain syndrome and wore orthotics for excessive pronation. Three-dimension analysis was performed using triplanar electrogoniomters to monitor the knee joint. Their objective was to determine whether soft orthotics affected the three-dimensional motion of the talo-crural/ subtalar joint and the knee joint in females with patellofemoral syndrome during both walking and running.

Gross and Foxworth (2003) wrote a commentary literature review on the role of orthotics as an intervention for patellofemoral pain. They reviewed four main points: the effects of orthotics on pain and function, the relationship between the foot and lower extremity patellofemoral joint mechanics, the effects of orthotics on the lower extremities and the effects of orthotics on patellofemoral joint position. Their findings indicated that during walking, there was a decrease in knee motion at initial contact and during midstance. This occurred in order to stabilize the foot for weight acceptance during the stance phase. Since orthotics were shown to reduce the movement of the foot, this in turn would reduce the movement of the knee motion in the medio-lateral direction. The reduction of knee motion during walking was due perhaps to a reduction of motion from the talocrural/subtalar joint, which reduced inclination at the knee.

An article that opposed the findings of Eng and Pierrnowski and Gross and Foxworth was that of Nester et al. (2003). They recorded kinematic and kinetic variables of walking with and without orthotics. The purpose of their study was to describe the effects of medially wedged and laterally wedged foot orthotics on the kinematics and joint moments of the rearfoot complex at the knee, hip and pelvis. In this study, subjects were asked to walk across a platform in each of the three conditions: shod (shoe), shod with a medially wedged foot orthotic, shod with a laterally wedged foot orthotic. Their results showed that orthotics had an effect on the rearfoot complex. Both medial and lateral wedged orthotics provided a more rigid characteristic to the lower limb, which therefore reduced the level of internal and external rotation at the ankle. By reducing the range of foot pronation to bring the foot more into a neutral position, the medially wedged orthotic was able to reduce the ability for the rearfoot to move in response to the internal rotation moment. However, authors reported no change in the knee joint motion due to the effects of the orthotics.

McCulloch et al. (1993) analyzed angular changes on the ankle and knee joints during stance resulting from wearing orthotics. Even at different walking speeds (0.889 and 1.333 m/s), authors found no differences in knee flexion, calcaneal eversion or subtalar joint pronation during the stance phase for any of the speeds tested. Speed influenced the angular changes at the knee during the stance phase, which led to maximum pronation, and increased dorsiflexion for a longer period of resupination after pronation had occurred. However, these changes occurred whether subjects wore orthotics or not. They concluded that orthotics did not reduce knee motion in the stance phase. The authors concluded that there would probably be a difference in knee motion if they had either increased the walking speed or looked at the differences between running and walking. Since they consist of two different techniques, there could be some possible changes between the two. This is possible since the two articles that studied both running and walking together (Eng and Pierrynowski, 1994 and Gross and Foxworth, 2003), found differences in knee joint motion.

Other authors examined the effects of orthotic wear in specific populations. Woodburn et al. (2003) investigated the kinematics of walking with orthotics in a population with rheumatoid arthritis. Their study included ninety-eight patients with bilateral arthritis of the peritalar complex and valgus deformity of the heel. Fifty patients were prescribed foot orthotics; the other forty-eight patients were in the control group. Kinematic data was obtained using an electromagnetic tracking system. They investigated how custom-made foot orthoses would affect the rearfoot valgus in these patients. Without the use of orthotics (control group), the rheumatoid arthritis patients were measured to have excessive subtalar joint eversion motion through the stance phase. Their findings showed that with orthotics, there were changes in the ankle-joint-complex. Also, with orthotic use, their subtalar joint motion was reduced and re-established balance since the foot was in a more neutral position. Overall, there were differences in the peak inversion and eversion periods from mid-stance to terminal stance. They concluded that orthotics were most useful during the loading response and terminal stance since that was when the ankle-joint-complex was most vulnerable to becoming unstable.

Orthotic devices have also shown to improve functionality for people who suffered from previous injuries. Tang et al. (2003) studied individuals who received reconstructive flap operations on their heel. Most patients with a reconstructed heel often do not make heel contact upon initial contact because of the lack of tissue under the calcaneus. As a post-operative treatment, they were all prescribed total contact insoles where there was a slight heel elevation to improve their walking ability. With their orthotics, patients were able to achieve heel-toe walking similar to that of healthy subjects. These studies suggest that there was less pronation at the rearfoot with orthotic use and orthotic use brought the foot back into a more neutral position for those with diagnosed pathologies (Woodburn et al., 2003, Tang et al., 2003). Even in walking, orthotics helped to correct excessive pronation in order to bring the foot back into a neutral position. Johanson et al. (1994) investigated different orthotic posting methods which included an unposted orthotic shell, forefoot post, rear-foot post and an orthotic that had both rear and forefoot posts. They wanted to determine what effects these postings had on controlling abnormal foot pronation during walking on a treadmill in individuals with forefoot varus deformities. These subjects were given four types of orthotics with a control condition to evaluate which of the four produced the most effective control on abnormal foot pronation. Authors concluded that rear-foot posting effectively controls subtalar joint pronation between heel strike and toe-off. In conclusion, both wedged orthotics (medial and lateral) reduced subtalar joint pronation between heel-strike and heel-off compared to without. This study demonstrated how orthotics helped subjects with rearfoot deformities in the stance phase and how they are most useful in controlling subtalar joint pronation in the stance phase. Overall, by evaluating the lower limb kinematics while wearing orthotics during walking, this will aid in inferring any differences in knee joint motion and better evaluating the differences between walking with orthotic wear or walking without orthotics.

2.4.3 EMG

Electromyography (EMG) is a method used to record muscle activation patterns in order to explain the performance of specific muscles in everyday activities such as walking. Tomaro and Burdett (1993) investigated the muscle activation patterns during walking for three muscles of the lower leg (tibialis anterior, peroneus longus and gastrocnemius muscles). The authors concluded that the activation of the tibialis anterior (TA) was prolonged with orthotic use over a longer duration following heel strike in the walking cycle. Authors interpreted their findings to reflect that TA functions to control the loading of the forefoot during stance. The authors suggested that the tibialis anterior might display higher EMG activity because there was more demand on that muscle to control the rapid descent of the forefoot immediately after heel contact. The elevated activity of the tibialis anterior may also be linked to its role in resupination of the foot in preparation for terminal stance since many studies have reported a lower muscle activity in maximum pronation with the use of orthotics (Woodburn et al., 2003, Tang et al., 2003, Nester et al., 2003). Overall, the consensus remained that the changes in muscle activity mainly affected the muscles closer to the foot than the muscles surrounding other joints while walking with orthotics.

2.5 Orthotics and running

2.5.1 Kinetics

Force plates are typically used to measure vertical, medio-lateral and anterio-posterior ground reaction forces (GRFs) during locomotion. From the studies investigating these biomechanical parameters, the majority has concluded that running with orthotics decreased the impact forces upon heel contact. Perry and Lafortune (1995) studied the impact loading during running and walking to investigate the effects of increased and decreased foot pronation. In their study, shoes were modified using valgus, varus and normal insoles. The varus footwear produced the highest resultant GRF of all three conditions during running. The authors concluded that impact loading was increased when normal pronation was restricted during running. However, no other studies have validated these findings or gone a step further in investigating the effects of orthotic wear on kinetic characteristics during running.

2.5.2 Kinematics

The use of orthotic intervention is a regular occurrence in running related activities. Some orthotics are used in medial or lateral posting to bring the foot to a more neutral position. Stackhouse et al. (2004) studied fifteen subjects who ran with either rearfoot or forefoot strike patterns, with no history of orthotic use. The purpose of their study was to compare the different effects of custom-made orthoses (medial and lateral post) on the lower extremity mechanics of the forefoot and rearfoot strike patterns while running. The results showed that there were no overall effects of orthotic wear on the forefoot or rearfoot strike patterns. Some subjects demonstrated a reduction in rearfoot eversion with orthotic use, suggesting that the response to orthotic use may be subject-specific. The same study also showed that knee adduction/abduction angles decreased with orthotics. This finding was in accordance with the findings of Eng and Pierrynowski (1994), who also showed that in both healthy persons and subjects who required orthotics, knee medio-lateral motion was reduced. In addition, these researchers concluded that there were no differences between different orthotic conditions.

Stacoff et al. (2000), studied the effects of medial foot orthoses during the stance phase, agreed with all the studies previously mentioned, in that they did not observe an effect of orthotic wear on kinematic patterns during running. They all conclude that while orthotics may be subject-specific, overall they do show some reduction in the medio-lateral motion of the knee. Bates et al. (1979) found a decrease in the period of pronation and the amount of maximum pronation in runners who wore orthotics. The authors observed that increased running speeds might have produced increased pronation as changes in leg orientation occurred. Their study confirmed that orthotic use could modify different aspects (such as maximum/ minimum pronation) of lower limb mechanics during the stance phase. Perry and Lafortune (1995) studied ten male subjects who ran on an 18-meter runway and wore regular running shoes, shoes with a valgus (medial) wedged insole and shoes with a varus (lateral) wedged insole. The purpose of their study was to determine the changes in impact loading induced by modifications in the amount of pronation of the foot. Overall, the magnitude of impact was affected by pronation modification. In other words, depending on the type of orthotic, it was possible to alter the amount of impact forces on the ground. During running, when the impact forces increased, there was a decrease in pronation. However, there were no changes in impact forces when pronation increased. Under the varus condition, pronation was decreased while pronation increased with the valgus orthotic. Additionally, there were changes in the rearfoot angles between the two types of orthotics. The valgus (lateral) wedged orthotic reduced the maximum rearfoot angle compared to the varus (medial) wedged orthotic and the no orthotic condition. Overall, the authors conclude that orthotics help to control the amount of pronation. Finally, in a study by Stackhouse et al. (2004), the characteristics of motion of the lower leg were recorded while healthy subjects ran with or without orthotics. Their main purpose was to compare the different effect of custom orthotics on the lower extremity mechanics of the forefoot and rearfoot strike patterns. This study consisted of fifteen runners with no history of orthotic use. All the subjects were categorized as rearfoot strikers. In the frontal plane, the knee adduction/abduction angles showed a decrease with orthotics. With orthotics, there was also a decrease in peak knee flexion, knee flexion velocity and knee flexion excursion with orthotic use. This finding was in accordance with that of Eng and Pierrynowski (1994), who tested foot orthotics in the frontal plane on adolescents with patellofemoral syndrome. They concluded that with orthotics, the knee medio-lateral displacement was reduced during the contact and mid-stance phase of walking but was increased during the contact and mid-stance phases of running in subjects with patellofemoral syndrome who were prescribed orthotics. Overall, the literature indicates contrasting findings in regards to the effect of orthotic wear on kinematic characteristics during running, which may be dependent on the population studied (healthy, pathological), as well as being subject-dependent.

2.5.3 EMG

Various studies have indicated that orthotic wear affects walking and running patterns of muscle activation. A study by Mundermann et al. (2003) recorded muscle activity in the lower extremities during the stance phase of running using three different kinds of orthotics (posting alone, molding alone and posting and molding orthotic) and a control condition. The muscles being studied included the tibialis anterior, peroneus longus, gastrocnemius, biceps femoris, vastus lateralis, rectus femoris and vastus medialis. The EMG aspects of this study helped to determine what muscles were activated during the different phases of locomotion, in particular the stance phase. Muscle activity was analyzed and averaged over 50 ms before heel-strike, 50 ms after heel-strike and early stance. The study showed that with the help of orthotics, there was an increase in the activation pattern of the tibialis anterior muscle both at pre and post heel strike. In addition, the posting condition caused a change in the mechanics of the lower extremities during the stance phase of running. In general, muscle activity was more decreased under molding conditions than posting and posting and molding conditions. In a similar study, Nawoczenski et al. (1999) recorded from superficial muscles of the lower leg, including the tibialis anterior, medial gastrocnemius, vastus medialis, vastus lateralis and the biceps femoris using surface electrodes. They found that with custom-made orthotics there was also a higher EMG activity of the tibialis anterior muscle, during the first 50% of the stance phase of running. In contrast, with orthotic wear, the biceps femoris muscle showed a lower EMG muscle activation during the stance period. The authors hypothesized that this occurred probably because of the reduced need for controlling internal tibial rotation (lateral hamstring muscle responsible for internal tibial rotation). From these studies, it is evident that the tibialis anterior muscle activity is likely affected by the use of orthotics. Since this muscle acts as an agonist of ankle dorsiflexor and foot invertor, it would seem likely that orthotic wear would influence the recruitment patterns of this muscle. However, the influence of orthotic wear on the activity of this and other muscles of the legs is still not clear and needs to be further investigated. By determining the activation pattern of certain muscles during running, we will be able to evaluate which muscles are mostly affected by orthotic use.

2.6 Flat feet and gait

Pes planus is a term described as flat feet, which is due to a laxity of the medial longitudinal arch in the foot. This fallen arch is normally attributed to several factors such as excessive tension in the triceps surae, laxity in the plantar calcaneonavicular ligament or plantar fascia (Van Boerum and Sangeorzan, 1999). A study by Ledoux and Hillstrom (2002) examined vertical ground reaction forces on seven regions of the foot during walking in persons with flat feet. They concluded that there was a greater force under the subhallucal area (located underneath the great toe) for those with flat feet compared to those with 'normal' feet. Since the osseous structure of the foot might differ between those with flat feet and those without, authors of this study hypothesized that this may cause some altered distribution of pressure patterns throughout various regions under the foot. In turn, this may provoke various compensation patterns between muscles of the lower leg.

In a subsequent study, the same authors (Van Boerrum and Sangeorzan, 2003) further investigated the biomechanics and pathophysiology of flat feet. They concluded that the triceps surae muscle might play an important role in maintaining the medial longitudinal arch in conjunction with the ligaments and bony structures in the foot.

In a previous study, Gray and Basmajian (1968) had found that the peroneus longus muscle is most active during midstance and toe-off of normal gait and that subjects with flat feet tend to have a higher muscle activation of the peroneus longus muscle compared to subjects with 'normal' feet. Indeed, it has been demonstrated that the peroneus longus muscle remains active during toe-off in order to prevent excessive inversion of the foot (Ledoux and Hillstrom, 2004, Gray and Basmajian, 1968). These studies suggest that the shape of the foot, in particular whether the foot is evaluated as having a normal or a flat medial arch, may affect motion, forces and activity of the lower leg muscles during gait. Although these effects may be small, they suggest that foot shape may act as a confounding factor in studies of gait characteristics. Moreover, studies have yet to examine the combined effect of foot shape and orthotic wear on biomechanical characteristics of gait, which constitutes the goal of this study.

Overall, evaluating the effect of orthotics on some biomechanical characteristics such as the ones previously described will aide in evaluating the functional use of orthotics. It is thought that the shift from lateral to medial aspects of the foot during the stance phase will be more restricted since from the literature, it is said that orthotics help to reduce pronation. A verification of this hypothesis would help to support the findings that pronation is decreased with orthotic use. So far, most studies show conflicting kinetic, kinematic and EMG results which could in part be due to the various types of orthotics (medial wedged, lateral wedged, etc.) tested in each study. One variable that requires further research would be the ability to quantify inclination of the tibia. Tibial inclination in the frontal plane is a method of evaluating the functionality of foot orthotics. Quantifying tibial inclination will help in determining the range of the tibia with orthotics and without orthotics. By analyzing the inclination of the tibia in relation to the vertical axis, this can help to determine any changes with orthotic use. One hypothesis that can be tested would be since there is an increase in knee motion as the body goes from walking to running, will there be more tibia motion in order to accommodate an increase in weight bearing on the lower extremities? To date, there are no studies available to answer this question while dealing with the effect of foot orthotics and tibial inclination. Further research will help to cover all parameters on the benefits of orthotics. Moreover, by evaluating the differences between subjects with flat feet and subject without flat feet will provide further insight into how foot structure may affect these biomechanical parameters.

Chapter 3- OBJECTIVES

3. Stating the problem

3.1. Rational

To date, few studies have investigated the overall effects of custom-made orthotic wear on the kinetic, kinematic and electromyographic characteristics of both walking and running. Foot orthotics are commonly prescribed by physicians in an attempt to reduce the incidence of leg injuries that are thought to be attributed to foot structural problems. However, orthotic fitting is often assessed in static (non weight-bearing or quiet standing) conditions. The way that this structural remedy affects how individuals walk and run, which are situations where the body is dynamically subjected to high loads, has seldom been systematically investigated. Muscle activity, kinematics of the legs and ground reaction forces recorded during gait allow a quantitative analysis of the effects of orthotic use on the characteristics of walking and running. From this proposed study, we will attain a better understanding of how orthotic wear affects the running and walking gaits for individuals with and without flat feet. By studying these aforementioned components, there can be greater comprehension into the use and purpose of orthotics.

3.2 Purpose

The purpose of this study is to determine any changes in the kinetic, kinematic or electromyographic characteristics of running and walking while individuals with and without flat feet wear custom-made foot orthotics. This study will help to predict the functional characteristics of the role of orthotics during running and walking. Present studies indicate differences in orthotic use but they have yet to be determined using these three biomechanical assessment techniques under these two conditions (walking and running). Research into this field will contribute to further knowledge in the study of human walking with orthotics, which may lead to better prescription of orthotics and insight into orthotic use and its biomechanical effects and functions.

3.3 Hypotheses

Our working hypotheses are that using our state-of-the-art biomechanical research facilities, we will be able to detect subtle changes occurring at the lower leg as a result of walking and running with orthotics. Moreover, in regards to the nature of our expected findings, according to the literature review, there has not been a general conclusion as to the effectiveness of orthotic wear while running or walking. Some studies have concluded that there are significant changes with orthotic intervention while others have recorded that those same changes were minimal or not significant. The parameters that will be tested are the same parameters also tested in the majority of these studies, with the addition of others that will constitute a more complete study. From the literature, there was an effect of orthotic intervention on electromyography (EMG) of the tibialis anterior muscle in both running and walking. In running alone, there was an effect on the biceps femoris muscle as well. Other parameters such as knee, ankle joint motion and ground reaction forces have been inconclusive in regards to orthotic wear.

According to our own hypotheses, with orthotics, we expect to see a decrease in tibial inclination in both running and walking because of the hypothesized ability of the orthotic to restrict the medio-lateral movement of the foot during gait. Since orthotics are known to improve the alignment of the joints of the lower limb, we hypothesize that there will be less movement at the knee and ankle joint which will cause less movement of the shank. We predict a decrease in medio-lateral ground reaction force with the use of orthotics due to the restriction capabilities of the orthotic device. These anticipated outcomes will be tested and evaluated to determine if there are in fact any interactions between the dependent and independent variables. Finally, since the sample group will be divided into two groups (i.e. subjects with flat feet and subjects without flat feet), it is hypothesized that there will be a higher EMG activation level of the peroneus longus muscle and the triceps surae muscle group due to the structure of the flat foot compared to subjects that do not have flat feet.

3.4 Objectives of the Study

The general objective of this thesis is to better understand and quantify how orthotic wear affects the biomechanics of walking and running. The specific objectives are to:

1- Determine how much tibial displacement occurs during the stance phase of running and walking with and without orthotics;

2- Evaluate the peak ground reaction forces during walking and running, with and without orthotics;

3- Measure the effects of orthotic wear on the activity of the main leg muscles during running and walking;

4- Determine any differences in biomechanical parameters between flat feet and non-flat feet during either condition.

3.5 Limitations

There are a few points that are noted in this experiment that will bring some limitations to this project. Firstly, due to laboratory space limitations, we did not constrain a specific running speed in our experimental protocols. This is in contrast to other studies such as that of Mundermann et al. (2003), who used a running speed of 4 m/s. Other studies have used a treadmill to constrain their subjects' running or walking speeds. Another consequence of having a restrained laboratory space is that it is possible that subjects were not able to reach a natural running gait when arriving within the recording zone, so that the recorded movements may not accurately reflect the subject's natural running style. Since this study required a specific population (those who wore orthotics), the sample size was small. However, in comparison with other orthotic studies, this sample size adequately represented the average number of subjects used in previous orthotic studies.

3.6 Delimitations

Results of our study will only apply within the limits of our experimental design (i.e. self-selected running speed) and our sample characteristics. For example, it is possible that the gait styles of our subjects were affected by the limited run-in space in the laboratory environment and the constraint of applying the entire foot on each platform. To minimize the impact of this factor, we were able to modify the placement of the force plates within the ground in order to adapt to each subject's step length. Additionally, the results from this study will be related to the floor surface of the laboratory or a surface similar to that of the laboratory Another delimitation of our study concerns the reasons why orthotics were prescribed to each subject initially. The purpose of our study is not to infer correlations between various injury types and severity levels and the biomechanics of running or walking but simply to quantify the mechanical effects of orthotic wear on the characteristics of gait. Finally, we deferred to physician referral in order to categorize our subjects as having either flat or normal shaped feet. Therefore, the results of our study will be generalized to a population of custom-made orthotic wearers in general.

3.7. Independent (IV), Dependent (DV) and Categorical (CV) variables

Independent variables

Independent variables are variables in an experiment that are being manipulated (Thomas and Nelson, 2001). The experimental or treatment variables, also known as the cause for this study were divided into two groups: with orthotics and without orthotics. The independent variables will help to determine any biomechanical changes in orthotic wear that occur as a result of analyzing the previously described objectives (see section 3.4).

Dependent variables

Dependent variables are used to measure the effect of the independent variable. In this case subjects with flat feet and subjects without flat feet were tested and compared using the following dependent variables. Also known as the yield or the effect variable (Thomas and Nelson, 2001), the dependent variables for this study are the changes in:

1) Peak muscle activation
2) Tibial inclination about the vertical axis
3) Medio-lateral ground reaction forces

under two given conditions (walking and running). All these variables were compared with respect to two independent variables and were also evaluated to determine if there were any interactions between them.

Categorical variables

The categorical variable also known as the moderator variable (Thomas and Nelson, 2001), is a type of independent variable however, this variable cannot be manipulated. These independent variables are not influenced by any of the two orthotic conditions. These variables are used to determine any interaction between the dependent and independent variable. For this study, the categorical variable is foot shape (flat feet versus 'normal' shaped feet). Together these variables (dependent, independent and categorical) will assist in determining the biomechanical changes under two set conditions (walking and running).

Chapter 4- METHODOLOGY

4. Methodology

4.1 Subjects

For this study, 14 subjects with no known lower limb injuries at least six months prior to testing were chosen. Eight subjects wore custom-made foot orthotics prescribed by a health care professional due to suspected excessive pronation (estimated by a diagnostic of flat feet in static, standing posture). The remaining 6 subjects wore orthotics for reasons other than excessive pronation. These participants were prescribed orthotics for either having tibial stress fractures, high arches or plantar fascitis. All subjects' orthotics were prescribed by a health care professional. Out of the 14 subjects (see table 4.1), 11 subjects trained regularly, (a trained subject refers to someone who works out between 2 to 4 times per week or plays recreational sports at least once per week). In the group with flat feet, the mean age was 27.5 years; mean height: 168.1 cm; and the mean body mass: 66.93 kg. The group without flat feet had the following averages: mean age: 30.7 years; mean height: 170.6 cm; and the mean body mass: 67.4 kg.

Table 4.1 Summary of subject demographics

	Age	Sex	trained	Height (cm)	Body mass (kg)	Flatfeet
	26	M	yes	172	70.5	yes
	26	F	no	159	49.0	no
	27	M	yes	170.2	63.5	yes
	23	F	yes	157.5	56.7	yes
	21	F	yes	165.1	49.9	yes
	15	F	yes	167	49.9	yes
	34	M	yes	172.7	79.4	yes
	27	F	no	172	61	no
	51	M	no	167.6	102.1	yes
	23	M	yes	173	63.5	yes
	23	F	yes	170.2	61.2	no
	66	M	yes	177.8	82.1	no
	24	F	yes	177	91	no
	18	F	yes	168	60.2	no
Average:	28.9			169.2	67.1	

This research project was approved by the Ethics Board of the Centre de recherche interdisciplinaire en réadaptation du Montréal métropolitain (CRIR), and each subject signed informed consent forms upon their arrival at the laboratory for the experimental session. Each subject also filled a PAR-Q questionnaire. The experiment took place at the Research Center of the Jewish Rehabilitation Hospital in Laval, Quebec.

4.2 Equipment

Data collection consisted of the synchronization of three data acquisition systems: a motion analysis system, a system of three force plates and an EMG telemetric system. Data was concomitantly recorded from these systems using a desktop computer, which was part of the laboratory set up (figure 4.1).

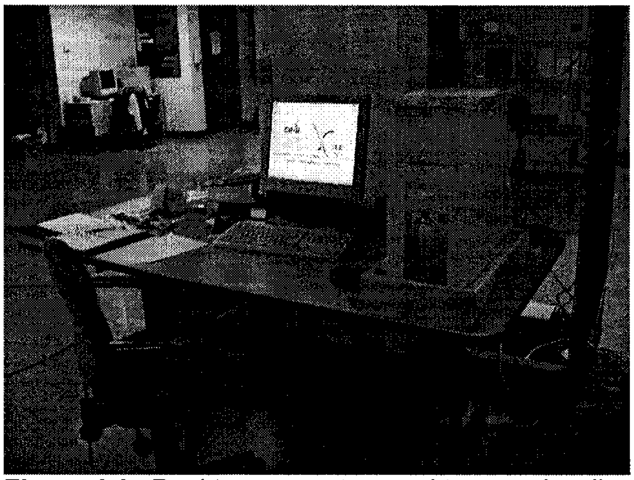


Figure 4.1: Desktop computer used to record, collect and analyze data from all three data acquisition systems.

A trigger from the computer software was used to begin the recording of each trial. Data of all three systems were synchronized and recorded using the VICON© Workstation™ software.

4.2.1 Forceplates

The integration of a system of three force platforms (figure 4.2) allowed the calculation of the ground reaction forces that were exerted by the foot during walking and running. Ground reaction force data was collected using AMTI© force plates (OR6-7, Scottsdale, AZ) that measured three force and three moment components along the x, y and z- axes.

Figure 4.2: The 10- meter walkway which includes three embedded AMTI© force plates.

Data from the force plates was recorded at a sampling frequency of 1080 Hz using the VICON workstation data acquisition software.

4.2.2 Motion analysis system

The VICON© system is an automated motion capture analysis system that tracks the position of passive reflective markers in space (figure 4.3). Marker displacement was tracked in 3D space and recorded during the protocol using near infrared light emitting diodes and 6 high-speed cameras. The sampling frequency of the VICON© cameras was set to 120Hz.

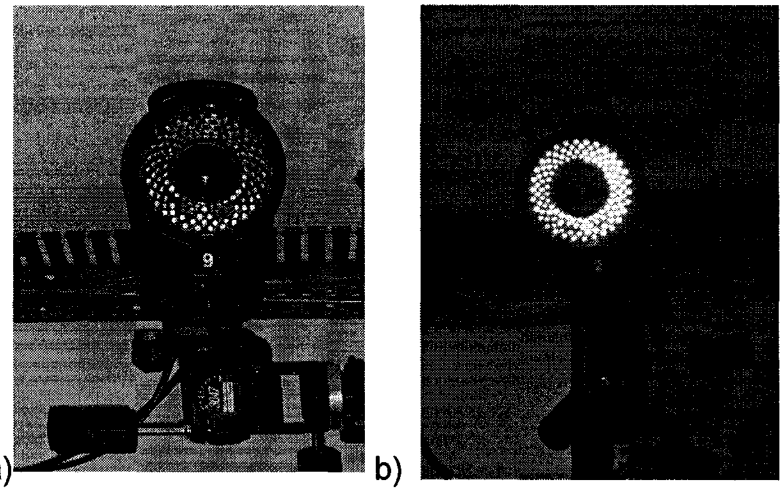

a) b)

Figure 4.3 VICON © cameras: One of six VICON © motion analysis cameras placed around the laboratory to capture the movement of the subject.

4.2.3. Telemetric electromyography system

To detect muscle activity levels, silver-silver chloride (Ag/AgCl) surface electrodes (Ambu© Blue Sensor SP, King City, Ont.) were used in the experiment (figure 4.5). Surface electrode disks had a length of 42.5mm x a width of 38mm (www.ambu.com). The electrodes used in this protocol are coated with an insulating foam material on the outer surface to prevent leaking of the electrical signals. The telemetric amplifying electromyographic system is used to amplify and record signals from muscles. The surface electrodes were placed near the muscle belly, avoiding the motor points of each muscle. The leads were then connected to a Noraxon© telemetric pack (figure 4.4) which was placed around the subject's waist while they performed the experiment. Signals captured by the surface electrodes were then transmitted telemetrically to an amplifying system (Noraxon©, Scottsdale, AZ) and sampled at 1080 Hz, obeying anti-aliasing laws.

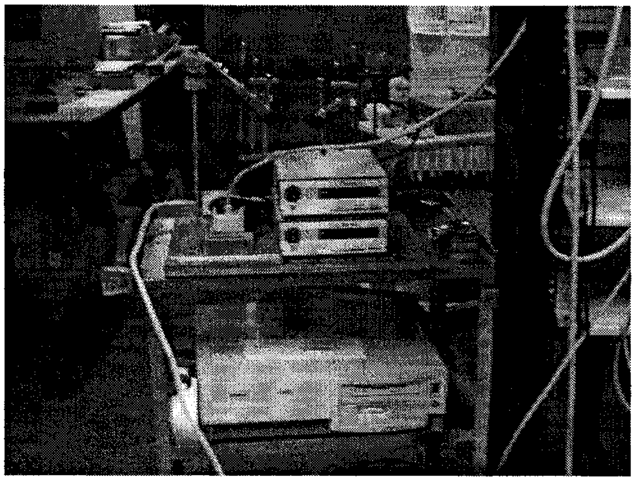
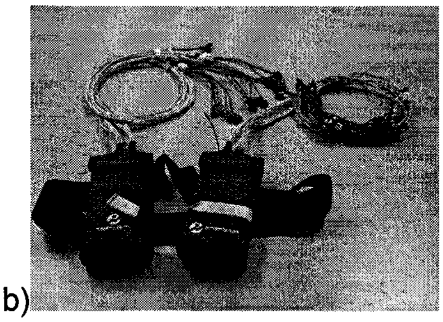
a)  b) 

Figure 4.4: a) EMG telemetric amplifying system (Noraxon©, Scottsdale, AZ) connected to the VICON © workstation; b) portable Noraxon© telemetric waist pack used to record EMG activity of 16 muscles.

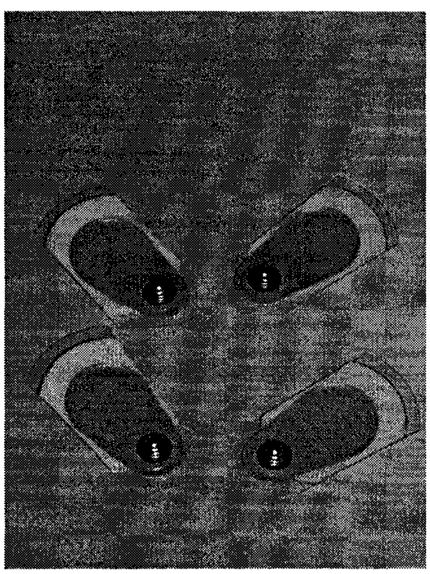
a)  b) 

Figure 4.5: a) Silver-silver chloride bipolar surface electrodes (Ambu ©, King City, Ont.) b) portable Noraxon© telemetric waist pack attached to two electrodes which are then attached to the belly of a superficial muscle.

4.3 Protocol

Upon subject arrival, the protocol was explained and subjects signed the informed consent form. Subjects were then weighed and their heights were measured. Afterwards, the bipolar, silver-silver chloride surface electrodes were placed bilaterally on the bellies along the fiber direction of the following leg muscles: tibialis anterior (TA), peroneous longus (PL), medial (MGAS) and lateral (LGAS) gastrocnemius, soleus (SOL), and the vastus lateralis (VLAT) and medialis (VMED) and biceps femoris (BFEM). Care was taken to obtain good quality signals by performing standard skin preparation procedures (cleaning, shaving, lightly abrading). Stabilizing the electrodes using adhesive bands insured minimal electrode movement on the skin surface. Muscle sites were palpated in order to isolate each one. To find each muscle, the actions needed are as follows (table 4.2):

Table 4.2 Direction of effort required to identify each muscle tested.

Muscle	Direction of effort
Tibialis anterior	Dorsiflexion
Peroneus longus	Eversion + plantarflexion
Medial gastrocnemius	Plantarflexion
Lateral Gastrocnemius	Plantarflexion
Soleus	Plantarflexion
Vastus lateralis	Knee extension + external rotation
Vastus medialis	Knee extension + internal rotation
Biceps femoris	Knee flexion

The left and right markers for the anterior superior iliac spine were placed medially and directly over the anterior aspect of the superior iliac spine. The markers of the left and right posterior iliac spine were placed laterally of the sacro-iliac joint. They were placed approximately where the spine joins the pelvis. The sacrum marker was placed at the tip of the sacrum. The left and right thigh markers were placed over the lower one-third of the thigh, laterally on the femur just below the hand when the arms are placed at the side of the body. The left and right knee markers were placed approximately where the knee joint axes pass through the lateral side of the knee joint. The left and right tibia markers were positioned laterally in alignment with the other markers of the lower leg, over the lower one-third of the shank. The left and right ankle markers were placed over the lateral malleolus. The left and right toe markers were placed over the head of the fifth metatarsal. Lastly, the right and left heel markers were placed on the posterior aspect of the calcaneus at the same level of the toe markers. This marker placement was based on the PlugInGait ™ kinematic model of the VICON© motion analysis system. The complete experimental set up including the portable Noraxon© telemetric waist pack is illustrated in Figure 4.6.

Figure 4.6: An example of the complete experimental set-up.

4.3.1 Walking trials

The velocity for the walking trials was at a self-selected comfortable walking speed. Subjects were instructed to walk across three force platforms. We accepted trials when the subject's foot made complete contact within each of the three force platforms. Subjects were allowed practice trials in order to practice making complete contact on each of the force platforms. This condition was repeated for 5 trials in each of the orthotic and non-orthotic conditions, with approximately 1-minute rest between trials. In order to account for the possible effects of habituation and fatigue on the data, the orthotic and no-orthotic conditions were presented in random order (i.e. either the first half of trials were accomplished with orthotics, or they were without orthotics). Furthermore, within each of the orthotic and no-orthotics blocks, the running and walking trials were randomized. In the walking trials, data from the three consecutive steps performed over the three force plates constituted the experimental data that was further analyzed. In running trials, data from the two steps where the foot landed on platforms 1 and 3 was analyzed.

4.3.2 Running trials

Subjects were instructed to run at a self-selected speed with and without their orthotics. In each trial, subjects ran across a 10-metre distance during which biomechanical data was recorded. To ensure that full heel-toe contact occurred on at least two of the three platforms (platforms 1 and 3), the subjects were allowed a few practice trials prior to actual recording. The subjects ran during 5 trials of approximately 5 seconds in each of the orthotics and no-orthotics conditions. Trials in each condition were performed sequentially with a rest period of approximately one minute between trials.

4.4 Data Analysis

The stance phase for the left and right sides were determined using kinetic data of each trial. This was obtained through VICON© workstation, which automatically identifies gait events (left toe-off, left foot strike, right toe-off, and right foot strike) according to the initial contact and toe off from the force plates throughout each trial. Data of one step per trial on either side of the leg (left or right) was extracted. Data from the five trials from each condition was further compared for each subject on both sides in order to confirm that there was no clear asymmetrical movement or muscle imbalance between both legs for each subject.

4.4.1 Kinetics

Data of the stance phases of steps performed over the three force plates were extracted using kinematic data. Analysis of the ground reaction forces (GRFs) were performed over three consecutive steps for each walking trial (two consecutive steps for running trials). Kinetic data was filtered using a low pass filter with a cut-off frequency of 6 Hz. Data processing was performed through Matlab™ custom routines. The peak range values for the medio-lateral GRF were obtained using the formula below:

$$\text{peak range} = \text{maximum} - \text{minimum} \tag{Eq. 4.1}$$

Data was then inputted onto an Excel™ spreadsheet. The average and standard deviation for each subject under each condition were calculated.

4.4.2 Kinematics

The VICON© Motion Analysis software was used to reconstruct the 3-dimensional leg movements. The data analysis software for the VICON© system, Workstation, was used to compile the data. Analyses of kinematic data of the knee marker were performed by filtering the raw data using a low pass filter, with a cut-off frequency of 6Hz. Two kinematic variables were extracted for data analyses. Firstly, the medio-lateral range of the knee marker displacement during one stance phase on the left and right leg. The second kinematic variable for analysis was the range of medio-lateral tibial movement. Angular displacement was obtained in the y-direction, in this case, the frontal plane.

Tibial inclination was obtained by finding the peak lateral and the peak medial positions of the knee marker during each stance phase for both left and right legs. The tibial segment was reconstructed with the position of the lateral knee marker and the ankle marker in space in the frontal plane. The range of tibial displacement was obtained by computing the difference between the peak medial and lateral tibial inclination angles. Using the filtered data, the range was obtained for each trial and each subject. The average range and standard deviation for each subject was then computed. An ANOVA model was used to identify any significant effects between each condition.

4.4.3. EMG

The peak activity values of each recorded muscle were extracted during the stance phases. EMG activity was analyzed from raw data collected from the first left and right stance phases from each trial. Each trial was filtered using a fourth order Butterworth filter with a bandpass frequency between 10- 350 Hz. The EMG waves were then full-wave rectified. Afterwards, the root-mean-squared (RMS) value of each muscle was calculated using Matlab™ data analysis software. The RMS formula was used to calculate the magnitude of each signal. This formula is as shown below:

$$x_{\mathrm{rms}} = \sqrt{\frac{1}{N}\sum_{i=1}^{N} x_i^2} = \sqrt{\frac{x_1^2 + x_2^2 + \cdots + x_N^2}{N}}$$

(Eq. 4.2)

where N is the total number of data points within the interval, x^2 is the value squared and Σ is the sum of all the squared values. These values were further analyzed in Microsoft© Excel™ where each trial was categorized into walking: with and without orthotics and running: with and without orthotics. The averages, standard deviations, percent standard deviations and the percent difference between conditions were calculated to compare the relative effect of orhotic wear on EMG activation for each condition (walking, running). The percent increase or decrease of EMG muscle activation was obtained by computing the difference in orthotic data and no-orthotic data divided by the control (no-orthotic) data, and mulitplied by 100:

$$\textbf{percent difference} = \left(\frac{\textbf{with - without}}{\textbf{without}}\right) * \textbf{100}$$

(Eq. 4.3)

These values are used to estimate whether there was a positive or negative change in EMG amplitude in each trial when the orthotic condition was introduced.

4.4.4 Statistics

We analyzed the data of the orthotic conditions in relation to data in the non-orthotic conditions for each subject. For the EMG data, the within-subject effects of orthotic wear was assessed by calculating the percent differences for each muscle (see above). Furthermore, kinetic, kinematic and EMG data were averaged across all subjects. In order to assess the overall effect of orthotic wear on all dependant variables, a two-way repeated measures ANOVA one factor model was used to identify any significant effects between each condition. For the ANOVA model, the significance level was set at alpha < 0.05. If there were any significant interactions, then a post-hoc analysis would be used to determine the location of the significant differences between the conditions.

Chapter 5- RESULTS

5. Results

Preliminary statistical analysis was performed to test the hypothesis that there is no difference between left and right side parameters. Paired T-tests revealed that this is true for all parameters except for the left and right peroneus longus, lateral gastrocnemius and biceps femoris muscles during the running condition. Therefore, we chose to present unilateral data for all parameters except those that indeed show a statistically significant bilateral difference. These analyses are presented in Appendix E. Furthermore; bilateral differences will be addressed in the Discussion section.

5.1 Walking

5.1.1 Kinetics

Below (figure 5.1) is an example of three medio-lateral ground reaction force (GRF) curves taken from subject 4 during a walking trial with orthotics. The medio-lateral forces were used in this study in order to compare how this shear ground reaction force component acts during both running and walking with and without orthotic intervention. The three force curves correspond to three consecutive steps recorded over three force plates used in the experimental set-up and represent a step made with the right foot, followed by one made with the left foot, and another one made by the right foot.

A positive force value indicates a ground reaction force directed towards the left in the absolute laboratory space, or a lateral ground reaction force for the left foot steps (second force curve) in the absolute laboratory space, or a medial ground reaction force for the right foot steps (first and third force curves). A negative force value indicates a lateral ground reaction force directed towards the right in the absolute space, or a lateral ground reaction force for the right footsteps and a medial ground reaction force for the left foot.

In figure 5.1, the initial negative peak of the ground reaction force indicates a lateral force applied on the right foot. From the first negative peak, the force curve becomes positive as the GRF is now applied in the medial direction on the right foot. This is consistent with the literature on gait kinetics (McGinnis, 1999). The initial positive peak of the second force curve indicates a GRF applied in the lateral direction on the left foot. This positive peak gradually progresses to a negative peak as the GRF progresses towards the medial direction. The third force curve exhibits the same pattern as the first GRF curve where initial contact was made with a lateral ground reaction force on the right foot, which progresses to a medial ground reaction force.

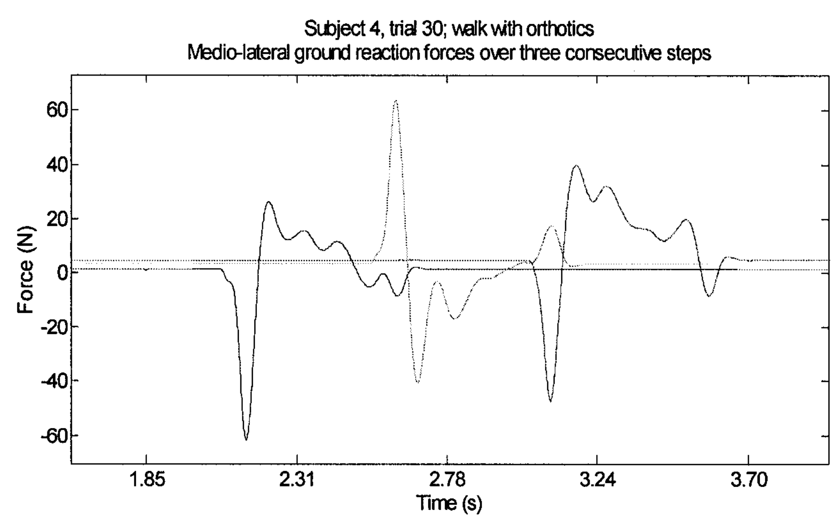


Figure 5.1: Example of medio-lateral ground reaction forces during walking for subject 4 with orthotics.

Figure 5.2 shows the differences between group average effects of orthotic wear on Fy during walking. Figure 5.2 shows the effect of orthotic wear in the medio-lateral direction for subjects with and without flat feet.

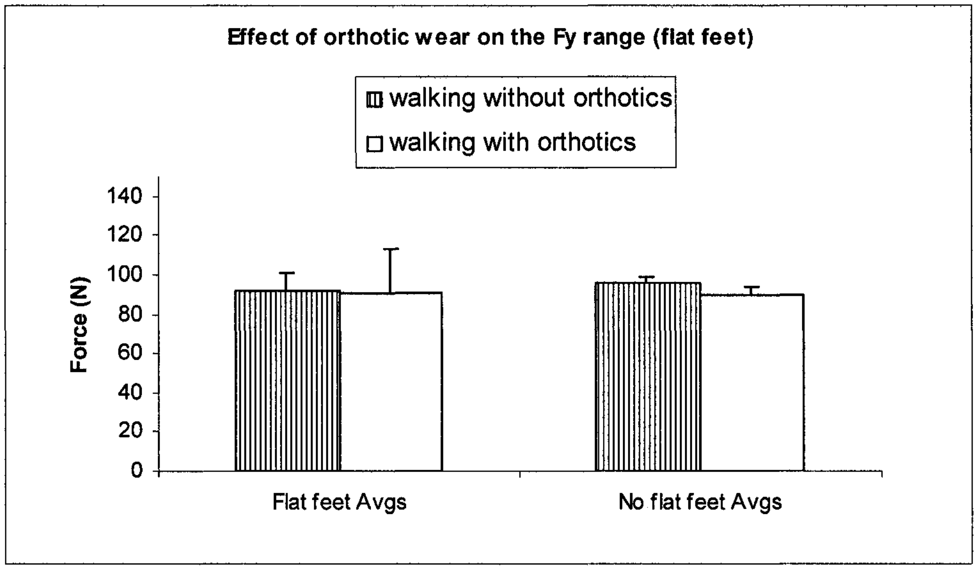


Figure 5.2: Effect of orthotic wear on the Fy (medio-lateral direction) for subjects with and without flat feet during walking.

The group averages for walking with orthotics were 90.42 +/- 8.13 N for subjects with flat feet, 89.39 +/- 9.46 N for subjects without flat feet. Walking without orthotics yielded a value of 92.31 +/- 5.75 N and 96.31 +/- 12.17 N for subjects with flat feet and without flat feet respectively. ANOVA revealed that there were no significant differences between walking with orthotics compared to walking without orthotics. The between group analysis revealed that there were no significant differences between groups of subjects with flat feet compared to subjects without flat feet.

5.1.2 Kinematics

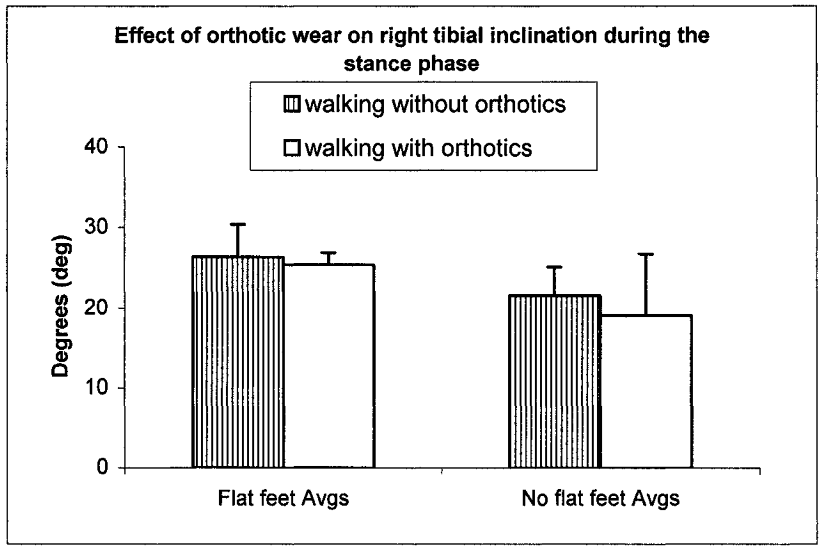


Figure 5.3: Tibial inclination during walking (group averages with and without flat feet).

Figure 5.3 is a bar graph representation of tibial inclination between both groups. Group average ranges for walking with orthotics are 25.38 +/- 1.94 degrees for subject with flat feet; 19.08 +/- 3.13 degrees for subjects without flat feet; and 22.68 +/- 2.34 degrees overall. ANOVA was performed and it was determined that there were no significant changes between either walking conditions as well as no significant changes were found between groups.

5.1.3 EMG

The following tables and figures give a representation of the raw EMG data before statistical analysis. Figures 5.4 to 5.7 display sample data (filtered) of 8 muscles (TA, PL, MGAS, LGAS, SOL, VMED, VLAT and BFEM). In subsequent analyses, these raw signals were rectified and the root-mean-square (RMS) values were used to calculate the area under the curve while the subject was in the stance phase. The stance phase is outlined by a solid black rectangle.

The main purpose of the TA muscle is to help control the fall of the center of mass (Nordin and Frankel, 2001). The TA muscle is activated during the first 15% of gait (figure 5.4) during swing and loading response.

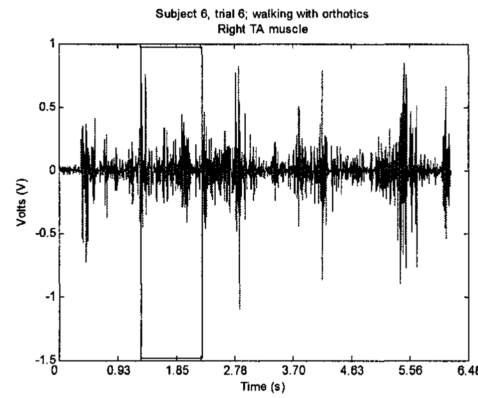


Figure 5.4: Example of filtered EMG burst of the right TA muscle during walking. The rectangle indicates the stance phase on the right side.

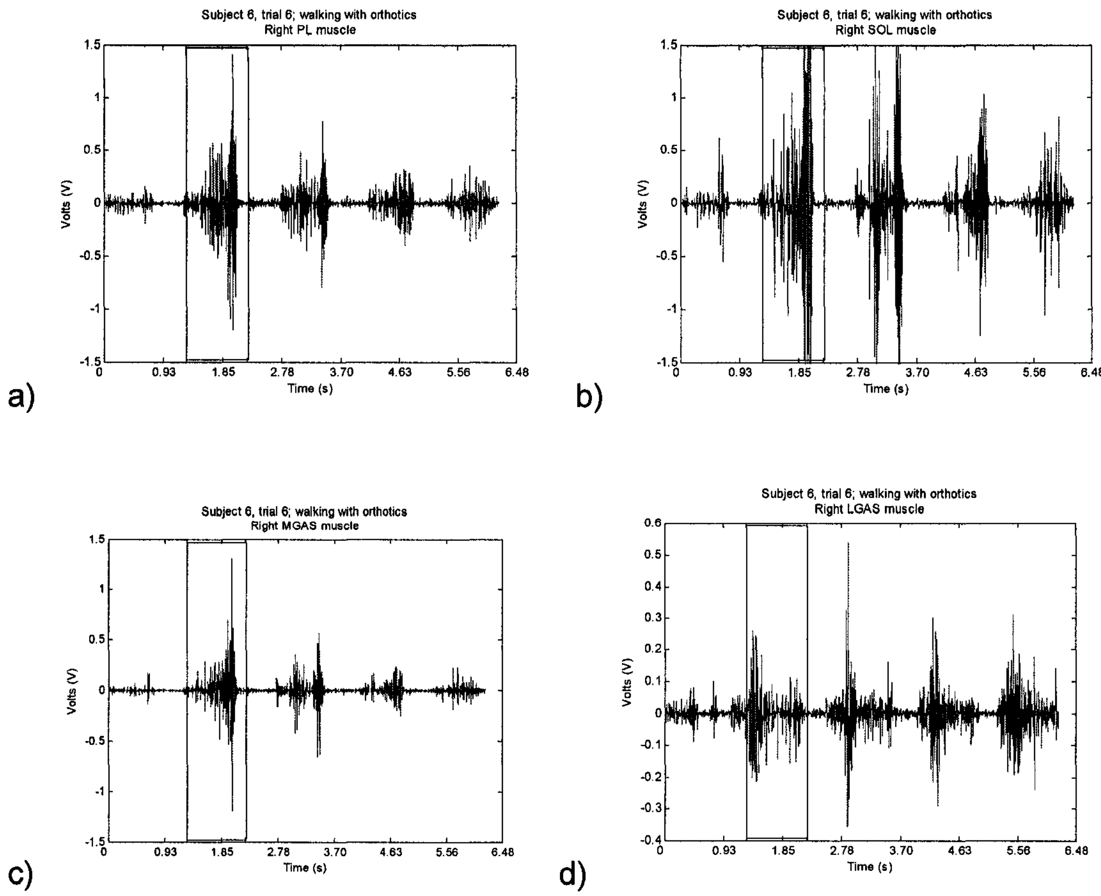
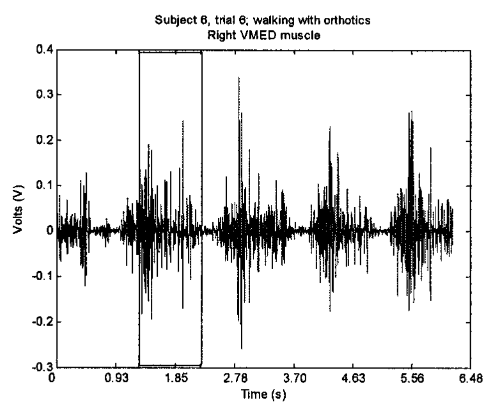
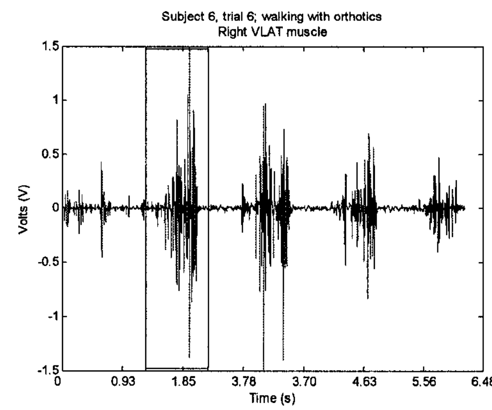


Figure 5.5: a) Example of filtered EMG burst of the right PL muscle during walking; b) Example of filtered EMG burst of the right SOL muscle during walking; c) Example of filtered EMG burst of the right MGAS muscle during walking; d) Example of filtered EMG burst of the right LGAS muscle during walking.

The PL muscle (figure 5.5a) is activated during early stance until terminal swing. The plantarflexor muscles, which include the PL, SOL, MGAS, LGAS muscles contract eccentrically during stance in order to control the advancement of the tibia, steady the knee joint, and to concentrically assist during toe-off (Nordin and Frankel, 2001). The SOL (figure 5.5b) and MGAS (figure 5.5c) muscles begin activation at 10% of the gait cycle. These muscles remain active throughout the stance phase up until pre-swing (Nordin and Frankel, 2001). The LGAS muscle (figure 5.5d) is activated during midstance. During terminal stance, the gastrocnemius muscles begin to plantarflex in order to stabilize the tibia. During the beginning of midstance to terminal stance the SOL, MGAS and LGAS muscles are activated (Nordin and Frankel, 2001).



a)



b)

Figure 5.6: a) Example of filtered EMG burst of the right VMED muscle during walking; b) Example of filtered EMG burst of the right VLAT muscle during walking.

The VLAT and VMED are activated during terminal stance. The quadriceps muscles (VLAT, VMED) help to control the knee as it prepares for weight acceptance and single limb support. These muscles contract to extend the knee through early midstance (Nordin and Frankel, 2001).

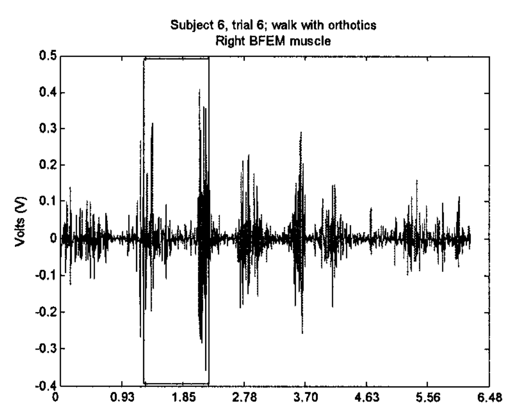


Figure 5.7: Example of filtered EMG burst of the right BFEM muscle during walking. The rectangle indicates the stance phase on the right side.

The BFEM muscle (part of the hamstring group) is activated in terminal swing (Nordin and Frankel, 2001). The BFEM helps to control and slow down angular acceleration of the knee as it goes into extension.

The following tables are group average calculations of the percent difference (% diff) of change (table 5.1) and p-values in the EMG RMS between the non-orthotic and the orthotic conditions obtained from the average EMG voltage values from each muscle in subjects during walking at right stance (table 5.2). The percent difference shows if there were any overall positive or negative changes for each muscle as a result of orthotic wear. A positive percent difference indicates a higher muscle activation while wearing orthotics. A negative percent difference indicates a lower muscle activation while wearing orthotics. A p-value less than 0.05 indicates a significant difference between the groups stated.

Table 5.1: EMG RMS percent difference (% diff) between walking and between groups during right stance.

(% diff) With vs. Without Orthotics		
Walking during right stance		
	Flat feet Subject Averages	**No flat feet Subject Averages**
TA	3.763	-16.90
PL	-7.222	0.947
SOL	14.09	-8.154
LGAS	11.90	2.451
MGAS	-12.22	-7.910
VMED	13.31	7.643
VLAT	16.74	8.836
BFEM	-1.768	2.059

Percent differences in bold indicate a significant difference.

Table 5.2: P-values (with vs. without, and flat feet and no flat feet) between walking and between groups during right stance.

(% diff) With vs. Without Orthotics		
Walking during right stance		
	P-Value **Walking With vs. Without Orthotics**	**P-Value** **Flat feet vs. No Flat Feet**
TA	0.255	0.762
PL	0.394	0.388
SOL	0.117	0.130
LGAS	0.694	0.0856
MGAS	0.512	0.118
VMED	0.217	0.263
VLAT	0.661	0.945
BFEM	0.576	0.867

P-values in bold indicate a significant difference.

Further to this analysis, we computed analyses of variance (ANOVA) on the EMG data. These statistical calculations revealed no significant effect of orthotic wear, for any of the muscles investigated, for the walking condition. Moreover, ANOVA comparing EMG patterns of subjects without flat feet to subjects with flat feet revealed no significant difference between both groups.

5.2 Running

5.2.1 Kinetics

Ground reaction forces during running are presented in figure 5.8 (below). Group averages of subjects with flat feet and subjects without flat feet are shown below.

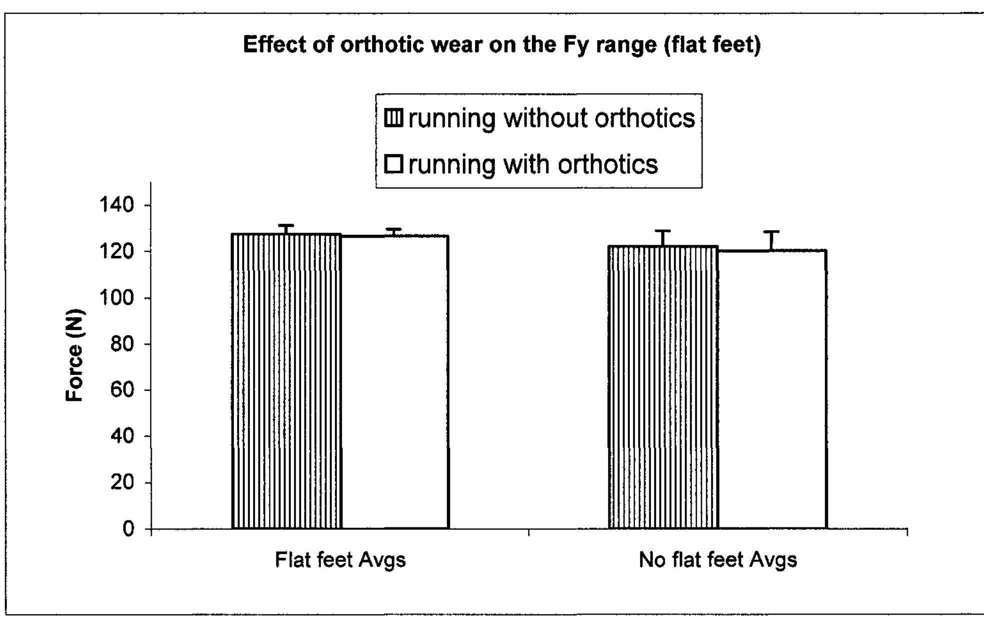


Figure 5.8: Effect of orthotic wear on the Fy (medio-lateral direction) for subjects with and without flat feet during running.

The averages for running with orthotics were 126.51 +/- 8.24 N for subjects with flat feet and 120.24 +/- 9.26 N for subjects without flat feet. Running without orthotics yields a value of 127.44 +/- 9.29 N and 122.13 +/- 13.14 N for subjects with flat feet, without flat feet respectively. ANOVA revealed that there were no significant differences between running with and without orthotics. In addition, there were no significant differences between groups (flat feet, non flat feet).

5.2.2 Kinematics

Figure 5.9 shows an example of tibial inclination throughout a running trial.

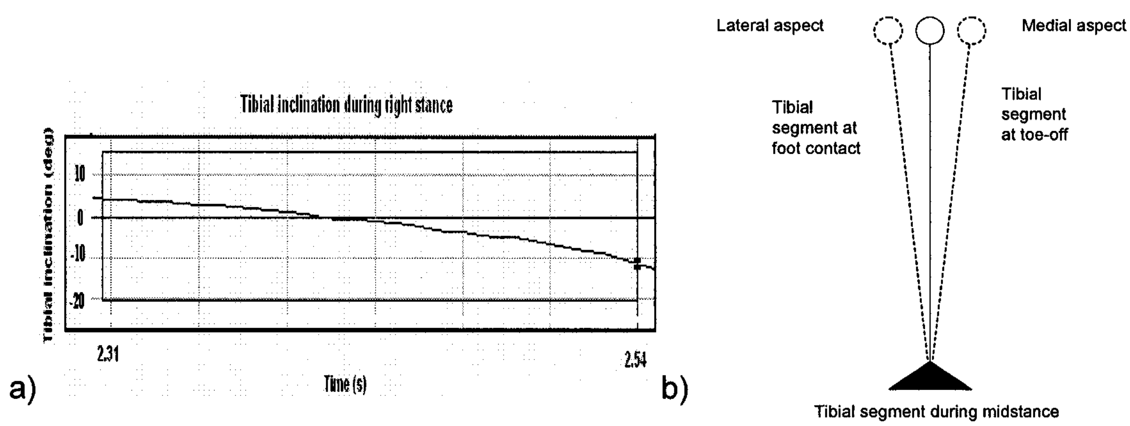


Figure 5.9: a) Angle versus time graph of right tibial inclination during the stance phase. b) Stick figure representation of the right tibial segment in the frontal plane (front view) during the stance phase.

In figure 5.9a, the tibial segment starts off at a positive angle, then subsequently passes through zero degrees around midstance and terminates in a negative angle value as it progresses from heel strike to toe off for the duration of the stance phase. Figure 5.9b is a stick figure representation of the tibial segment position throughout stance as seen in the frontal plane. Group averages for subjects with and without flat feet during both running trials are represented below in figure 5.10.

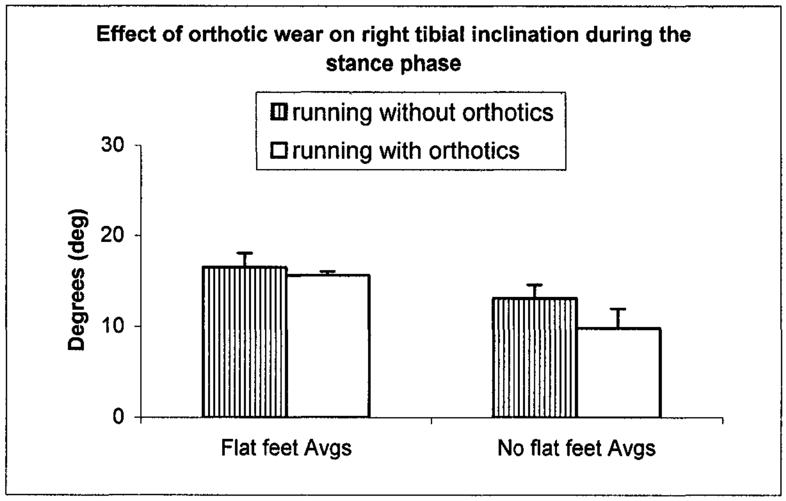


Figure 5.10: Tibial inclination during running (group averages with and without flat feet).

Group averages for running without orthotics were16.55 +/- 1.32 degrees for subjects with flat feet and 13.12 +/- 1.08 degrees for subjects without flat feet. Group averages for running with orthotics were15.64 +/- 0.77 degrees for subject with flat feet and 9.84 +/- 0.87 degrees for subjects without flat feet. ANOVA calculations did not yield a significant effect between running with and without orthotics. Moreover, there were no significant group differences (flat feet vs non flat feet).

5.2.3 EMG

Table 5.3 shows the EMG percent difference during the right stance of running. The muscles presented in this table are the muscles of the right side for group averages of subjects with and without flat feet. P-values are presented in the subsequent table (table 5.4), between orthotic conditions and between groups.

Table 5.3: EMG RMS percent difference (% diff) between running and between groups during right.

(% diff) With vs. Without Orthotics		
Running during right stance		
	Flat feet Subject Averages	**No flat feet Subject Averages**
TA	2.652	4.998
PL	6.080	9.607
SOL	**-3.944**	**-13.73**
LGAS	**-0.308**	**7.494**
MGAS	5.133	4.518
VMED	4.310	4.202
VLAT	8.309	-11.58
BFEM	4.557	-11.42

Percent differences in bold indicate a significant difference.

Table 5.4: P -values (with vs. without, and flat feet and no flat feet) between running and between groups during right.

(% diff) With vs. Without Orthotics		
Running during right stance		
	P-Values Running With vs. Without Orthotics	**P-Values Flat feet vs. No Flat Feet**
TA	0.304	0.265
PL	0.422	0.0959
SOL	**0.0187**	**0.0330**
LGAS	0.933	**0.0045**
MGAS	0.812	0.126
VMED	0.8058	0.318
VLAT	0.947	0.376
BFEM	0.961	0.750

P-values in bold indicate a significant difference.

ANOVA revealed there is a significant difference in SOL muscle activation with orthotic wear during running (p= 0.0187). Also, there are significant differences in the SOL and LGAS muscles (table 5.4) between groups with and without flat feet (p= 0.0330 and p= 0.0045 respectively).

Since there were significant differences in EMG activation between the left and right side during the stance phase for the PL, LGAS and BFEM muscles, table 5.5 represents the percent difference and p-values during the left stance during running for subjects with and without flat feet.

Table 5.5: EMG RMS percent difference (% diff) between running and between groups during left stance.

(% diff) With vs. Without Orthotics (flat feet)		
Running during right stance		
	Flat feet Subject Averages	**No flat feet Subject Averages**
PL	-2.502	3.778
LGAS	**-3.299**	**-11.52**
BFEM	-0.663	5.788

Percent differences in bold indicate a significant difference.

Table 5.6: P-values (with vs. without, and flat feet and no flat feet) between running and between groups during left stance.

(% diff) With vs. Without Orthotics (flat feet)		
Running during right stance		
	P-Values Running With vs. Without Orthotics	**P-Values Flat feet vs. No Flat Feet**
PL	0.569	0.099
LGAS	0.140	**0.0103**
BFEM	0.982	0.0834

P-values in bold indicate a significant difference.

Table 5.6 shows the p-values for the PL, LGAS and BFEM muscles on the left side. ANOVA confirmed that there was a significant difference between groups (flat feet vs. no flat feet) for the left LGAS muscle (p-value=0.0103).

Chapter 6- DISCUSSION

6. Discussion

6.1 General Hypothesis

The purpose of this study was to evaluate the biomechanical properties of orthotics during walking and running. Furthermore, the objectives of this project were mainly to:

1) Quantify the peak medio-lateral ground reaction forces during gait with and without orthotics.
2) Determine the amplitude of tibial inclination occurring during walking and running with and without orthotics.
3) Compare the muscle activity of the main lower extremity muscles during walking and running, with and without orthotics.

Based on the literature (Chapter 2), according to the literature, it was proposed that with orthotic wear, there might be a decrease in medio-lateral ground reaction forces during walking and an increase during running. This was not found in the present study; there were no significant differences in medio-lateral ground reaction forces between orthotic conditions for either walking or running.

Secondly, there were only equivocal data to predict the effect of orthotic wear on kinematic parameters of gait. Most studies had varying conclusions regarding kinematic data with respect to orthotic use in the frontal plane. This could be a result of the varying types of foot orthotics used in each study. From the results of the present study, there were no significant effects of orthotic wear on tibial inclination during either running or walking.

Lastly, the literature suggests that several muscles may be affected with orthotic use during either walking (tibialis anterior) or running (tibialis anterior and biceps femoris). Our results show that there were significant differences in the activation of the soleus muscle (p=0.0189) during running. Furthermore, there was a significant difference between both groups (flat feet vs. no flat feet) for the right and left lateral gastrocnemius muscles (p= 0.0454 and p= 0.01025 respectively). These findings will now be further discussed in the sections below.

6.2 Walking

6.2.1 Kinetics

In the present study, people with flat feet (n=8) and people without flat feet (n=6) were tested. Medio-lateral ground reaction forces (GRF) yielded no significant difference between orthotic wear during walking. Since walking is a fairly stable motion, it is possible that the GRFs were not affected by either orthotic condition. In a previous study, Nester et al. (2003) concluded that the amount of pronation affects shock absorption in the medio-lateral direction depending on the type of orthotic wedge used. In their study, they concluded that there was a decrease in medio-lateral GRFs using a laterally wedged foot orthotic. Therefore, authors hypothesized that the effect of orthotic wear on ground reaction forces may be dependent on the type of orthotic device. More specifically, a lateral wedge orthotic increases pronation and therefore reduces GRF in the medio-lateral direction during walking.

The absence of differences in medio-lateral ground reaction force in our study can be explained by the hypothesis that subjects with flat feet may instead require a medially wedged orthotic device since the majority of people with flat feet are categorized as having fallen arches on the medial aspect of their feet. However, the specific type of orthotics that these subjects wore was not categorized since they were custom-made. Moreover, our subjects without flat feet may have had various reasons as to why they wore prescribed orthotics and therefore the orthotic wedges, which were not quantitatively assessed in this study, may have varied also between these subjects. Therefore, our results could suggest that orthotics help to restrict movement in the medio-lateral direction during walking however, this is likely dependant on the degree of inclination from the wedge placed on the orthotic.

Another aspect to consider is that medio-lateral forces are the most variable of the three dimensions of GRFs. Moreover, these forces can differ between left and right feet of the same person (Craig et al, 1995 and Trew et al., 2001). Finally, another confounding factor could be that subjects were allowed to wear their own shoes during the experiment. Depending on the shape and stability characteristics of the shoe, more or less medio-lateral control may have also been provided by the shoe itself. In conclusion, according to the initial hypothesis, it was predicted that there would be a decrease in medio-lateral GRF with orthotic wear, since orthotics are known to restrict movement in the medio-lateral direction. However, the results of this parameter are not significant, so we cannot confidently state that there is a decrease in medio-lateral GRF due to orthotic wear during walking.

6.2.2 Kinematics

Since walking is a comparatively simple locomotor task, performed mostly in the sagittal plane, there is reason to expect limited movement to occur in the frontal plane, especially during the stance phase where the foot is constrained. For this reason, it is possible that there was indeed no significant deviation between walking with an orthotic device and walking without an orthotic device.

Overall, there were no differences in the degree of medio-lateral angular displacement of the tibia during the stance phase while subjects wore orthotics compared to without wearing orthotics. According to the literature, there had not been a previous study that had focused on medio-lateral deviation of the tibial segment during the stance phase of walking. However, common belief is that during walking the tibia is less prone to excessive pronation with orthotic use since one of its main purposes is to reduce the movement of the subtalar joint. The present findings do not support this hypothesis.

Some limitations associated with our study may also have affected our findings. In this experiment, speed was not set at one constant tempo; rather it was set at a variable self-selected speed determined by each subject. In a study by McCulloch et al. (2003), it was concluded that there were no changes in slower paced movements such as self-selected paced walking. McCulloch et al. (2003) tested subjects at 0.889 and 1.333 m/s. It was concluded that there were no changes in slower paced movements such as self-selected paced walking. Since it is commonly believed that walking at a natural pace does not cause any noticeable amount of tibial medio-lateral movement. In this experiment, there were no significant differences in medio-lateral tibial inclination during walking. During the experiment, since subjects were instructed to walk at a comfortable speed, they may not have walked at a speed that would initiate a movement that would cause excessive pronation. Rather, they likely remained within their own comfort zone and walked at a rate where there was perhaps no challenge from the subtalar joint. However, it is possible that had we tested different walking speeds, we could have identified differences between the tibial displacement values between orthotics and no orthotics conditions. Finally, although we qualitatively adjusted the placement of all three force plates within the floor to accommodate average step length, most subjects required several trials in order to insure placement of the feet onto the force plates. This adjustment could have also affected how naturally our subjects performed the walking task.

6.2.3 EMG

In previous studies, muscles of the upper leg have not been the primary area of EMG testing in regards to gait. The quadriceps muscles (vastus lateralis and medialis) and the biceps femoris muscles' main actions are to control the knee joint during terminal stance. However, this action occurs mostly in the sagittal plane. Since the muscles that are more distal to the subtalar joint act to stabilize the knee joint during the stance phase, the muscles most likely affected by orthotic wear would be those of the lower leg and ankle.

Based on the literature, we expected to observe that the tibialis anterior (TA) muscle (Tomaro and Burdett, 1993, Nawoczenski et al., 1999) and the peroneus longus (PL) muscle (Gray and Basmajian, 1968) would be affected by orthotic wear during walking. During walking, the tibialis anterior muscle was hypothesized by most authors as showing a significant change in muscle activation with orthotic wear since the tibialis anterior helps to control deceleration of the tibia at the beginning of the stance phase; however, the main role of the tibialis anterior is to control foot drop from terminal swing to heel-strike. In this study there was no significant impact of orthotic wear on the tibialis anterior.

6.3 Running

6.3.1 Kinetics

There was no significant difference in medio-lateral GRFs under both conditions (with versus without orthotics) for either group tested. From the proposed hypothesis, it was suggested that with orthotics, there would be a decrease in medio-lateral ground reaction forces during running and walking. A lack of significance in this part of our study may be due to the different types of orthotics used. Each subject required orthotics for different purposes. Our experimental group may likely have contained varying degrees of wedges as each subject in this group wore orthotics that fit their need and not necessarily orthotics that were all homogeneous in this experiment. Subject-to-subject natural variability and shoe type may have played parts in these findings.

6.3.2 Kinematics

The initial hypothesis in regards to the effect of orthotic wear on tibial inclination during running were derived from Gross and Foxworth (2003) who stated that during running, orthotics were found to restrict the ankle joint which would in turn restrict the deviation of the tibia during the stance phase. Another hypothesis from McCulloch et al. (1993) suggested that increasing the walking speed or analyzing different gait patterns such as running could highlight significant effects of orthotic wear.

For this reason, our original hypothesis stated that there would be a decrease in angular displacement of the tibial segment with orthotic wear based on the restriction of medio-lateral movement with an orthotic device. However, no results presented from the literature where found to support this hypothesis per se. Our study indeed showed no significant differences between the effects of orthotics on both groups. Our findings can be explained by the fact that medio-lateral shifting of the foot is variable from person to person (Craig and Oatis, 1995). Perhaps the shift in the medio-lateral direction varied amongst subjects and that these small shifts were undetected in our protocol.

This could be due to the subjects' individual running style or it can be sport-specific depending on whether the subject requires orthotics to play any sport that involves excessive medio-lateral movement of the lower limbs (i.e. basketball or tennis) or if the sport is more of a linear sport (i.e. long distance running) where the emphasis on medio-lateral shift is not as important. Stacoff (2000) concluded that orthotics do not have an overall impact kinematically; they may be subject-specific. Another explanation of our results is that also in running, there is limited medio-lateral movement during the stance phase. Since frontal plane motion during gait is minimal compared to the other planes of motion, the difference between orthotic and non-orthotic conditions may be small in absolute values and may not show any significance with tibial inclination in the frontal plane.

Other probable limitations in this experiment that may have affected these findings take into account that first, we did not control for running velocity (subjects were asked to jog at a comfortable pace). Also, due to the limited space available, subjects had only a few steps to accelerate to reach this comfortable pace. As a consequence, it is possible that subjects were still accelerating when they reached the force plates. Finally, subjects had to accommodate their step length to insure foot placement on the force plates, which required a few trials and could have masked their usual running style. Therefore, both the fact that subjects ran at their own preferred velocity and had limited space to accommodate and reach this velocity may have affected the results of our running trials.

6.3.3 EMG

A study by Nawoczenski et al. (1999) had previously concluded that there was a higher muscle activation of the tibialis anterior muscle and a lower muscle activition of the biceps femoris during running with orthotics. In addition, Van Boerum and Sangeorzan (1999) had concluded that the triceps surae muscle group plays a role in maintaining the medial longitudinal arch with the help of other structures around the foot. During running, it was hypothesized that the tibialis anterior muscle would be influenced by orthotic wear since it is an ankle dorsiflexor and foot invertor, which are functions that orthotics are thought to control.

Moreover, Nawoczenski et al. (1999) suggested there was a lower activation of the biceps femoris muscle since orthotics help to restrict internal movement of the tibia, a function in which the biceps femoris can be involved. In the current study, there was a lower activation of the soleus muscle with orthotic wear. Since the main action of the soleus muscle is plantarflexion of the ankle joint and assistance in steadying the leg over the foot, this significant effect in muscle activation could be attributed to the added restriction role of the soleus muscles in contributing to proximal internal tibial rotation with orthotic wear. Further investigation should focus on measuring internal tibial rotation in order to validate this interpretation.

Throughout our study, the percent difference served as a method to normalize the changes in EMG activation relative to the without-orthotics EMG RMS values of each muscle and each subject. The absence of significance in most of the EMG data could be due to the small sample size used or the accuracy in placement of surface electrodes for each subject. The nature of the surface EMG signal in itself may have posed as a source of error in this experiment. Common sources of error are the presence and the movement of skin and subcutaneous fatty tissue, which creates a barrier between the muscle signal and the surface electrode. Crosstalk between nearby or underlying muscles can also have an effect on the output of an EMG signal. Although expressing the effect of orthotic wear on EMG parameters using percent change reduces the impact of most of these factors, some source of error inherent to the surface EMG technique remains (e.g. electrode movement, electrode adherence changes throughout a protocol). Also, we chose to calculate our measures of EMG amplitude across a relative duration, the duration of the stance phase for each trial. Although this variability in the duration over which the signal is calculated may induce an absolute margin of error in the data calculated, our post-hoc analyses revealed that there was no statistically significant effect of orthotic wear on stance duration in either walking or running trials (0.677 +/- 0.062 s vs 0.684 +/- 0.078 s, $p = 0.27$ for walking without vs with orthotics; 0.322 +/- 0.041 s vs 0.315 +/- 0.038 s for running without vs with orthotics). Since our objective was to assess the effect of orthotic wear on EMG RMS parameters, if we consider that there are small (but insignificant) differences in stance duration, we can confidently say that the significant effects that we observe with respect to EMG RMS are indeed due to the EMG amplitude characteristics and not due to the duration over which these parameters are calculated. Another typical way of analyzing EMG data of a functional task could have been to calculate the integrated EMG from a low-pass filtered, rectified signal, average the linear envelopes for each muscle over several step cycles for each subject, and normalize the duration of each step before averaging across subjects. However, we feel that expressing EMG amplitude with respect to a functional event of the gait cycle (the stance phase) and not relative to an absolute time duration gives more functional meaning to the data assessed, i.e. the data represents the EMG amplitude as the limbs are in contact with the ground. This approach of expressing EMG amplitude relative to a functional time event is commonly used in studies of cyclical tasks (Balter et al., 2006; Bonato et al., 2001; Cote et al., 2005; Gerdle et al., 2000; Larsson et al., 1999; Oksa et al., 2002; Viitasalo et al., 1993).

6.4 Differences between the left and right sides

From our results, the activation levels on the left side for three muscles were higher than the muscle activation levels on the right side. The differences between the left and right sides for the peroneus longus, lateral gastrocnemius and the biceps femoris muscles could be due to the variability in muscle activation between subjects or the dominance of the left lower extremity strength over the right lower extremity in these particular subjects. This could also reflect slight differences in orthotic or shoe fabrication or simply accommodations to the force plate placements. The most important finding of this preliminary left-right similarity assessment is that in most parameters, there was no significant difference. This allowed us to limit our extensive analysis to parameters of the right side (for all parameters except the three mentioned above).

6.5 Effect of foot shape (flat feet vs. non flat feet)

In order to assess the possible confounding effect of foot shape on our findings, groups were divided into flat feet and non flat feet groups to determine if there were any biomechanical differences in foot structure between both conditions. During both walking and running conditions, our kinetic analyses did not reveal differences in medio-lateral ground reaction force patterns between both groups, whereas other studies have found differences in forces under specific regions of the foot between persons with flat feet and persons without (Ledoux and Hillstrom, 2002). This difference in findings is due to the fact that using our equipment; we were only able to quantify maximal force applied on the force plate during each step and not force patterns under isolated structures of the foot.

Also, under both tested conditions (walking and running), there were no significant differences in kinematic data between both groups, suggesting that people with flat feet who wear orthotics do not differ kinematically from those who wear orthotics in general. This could be due to the fact that overall, orthotics made for any condition (i.e. flat feet or shin splints) will not display differing effects on kinematic characteristics of gait (walking or running) in the frontal plane. Still, our methods were effective in identifying some effect of foot shape (see below), which most likely is compensated for by more proximal structures of the lower limb, resulting in unaffected tibial inclination in the frontal plane during gait (both walking and running).

During walking, there were no significant electromyographical differences between the effects of orthotic wear for either group. However, our findings on the effect of flatfeet during running are consistent with those from the literature in that we did observe differences in the activation of the soleus and lateral gastrocnemius muscles in both groups (flat feet, non flat feet). More precisely, there was a higher muscle activation for subjects with flat feet. This difference in muscle activation could be due to the differences in the supporting structures of the foot between those with flat feet and those without flat feet. Indeed, it has been previously suggested that people with flat feet have either a laxity in the ligaments of the foot to maintain an arch or too much force causing the foot to lose its arch (i.e. obesity).

6.6 Conclusions

The main purpose of this study was to examine the biomechanical effects of orthotic wear during walking and running. From the parameters tested in this experiment, one EMG variable during running has shown significance between orthotic and non-orthotic conditions as well as between groups. Some limitations of this study included allowing subjects to walk/ run at a self-selected speed. Since the rate of self-selected speed may vary from person to person, allowing subjects to walk/ run in a controlled environment freely causes variability between subjects. By setting set speeds, this could have decreased inter-subject variability and provided more homogeneous data. However, post-hoc analysis reveals that no significant differences were found between the stance phase durations of walking with and without orthotics, or between running with and without orthotics, suggesting that with respect to the objectives of this project, this limitation had a limited impact. Another limitation was a rather small sample size, and increasing the sample size would have possibly produced more reliable data. Another shortcoming of this research project included the difficulty in obtaining a homogeneous sample, in other words, being able to acquire a sample group that is as similar as possible (i.e. all males or the same pathology for all subjects in the normal group). Furthermore, acquiring orthotic devices fabricated by the same manufacturer for each subject tested could have also enhanced the homogeneity of the sample group. Finally, other instrumentation that could be used to attain better understanding of orthotic devices could include: indwelling EMG of muscles of the lower leg that have a direct action on the ankle joint, and recording gait biomechanics on different surfaces (i.e. on an incline, treadmill, or an indoor/ outdoor track). As well, a walkway of ~10 metres became a limitation in this study due to the restricted space each subject was allowed to walk or run.

The use of a larger laboratory space would have increased the number of steps taken in each trial and would have possibly allowed subjects to have reached a constant, comfortable gait speed once force plates were reached. Moreover, the constraint of having to place the feet on the force plates could have also slightly modified the natural gait style and induce some modification in the normal gait patterns. Another method to constrain gait speed as well as allowing the subject more freedom in foot placement could have been to use a force plate instrumented treadmill which can accurately measure speed for walking and running trials. Insole pressure sensors can be another method of quantifying the biomechanical changes with orthotic wear over more consecutive trials while not constraining subject to place the feet on precise locations on the floor. In particular, pressure sensors would allow us to quantify precise pressure patterns under specific regions of the feet during the stance phases of gait. With these supplementary suggestions, the effect of orthotics during running or walking gait can help uncover the answers towards the biomechanical benefits (advantages and disadvantages) related to orthotic wear.

6.7 Future Directions and Relevance

Studies on orthotics and their biomechanical advantages and disadvantages have not been extensively conducted to date. There are many conflicting conclusions from the literature as to the significant effects of orthotics during walking or running. Future directions involving research into orthotics can include looking at joint angles and moments of the ankle, knee and hip joints during locomotion. This can give researchers and clinicians a better understanding of the effects of orthotic wear on more general, combined measures of kinematics and kinetics of gait. With more studies into orthotic devices, there will be more information as to how and why there are an increased number of people who are prescribed orthotics in various environments. Overall, orthotics may have some potential advantages however further quantitative evidence needs to be provided in order to validate this hypothesis.

Chapter 7- References

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APPENDIX A
CERTIFICATE OF ETHICAL ACCEPTABILITY

APPENDIX B
RECRUITMENT FLYER (ENGLISH & FRENCH)

APPENDIX C
INDIVIDUAL SUBJECT PERCENT DIFFERENCE TABLES

The tables presented below are tables used to display the individual EMG subject percent difference.

Table C.1: EMG percent difference (% diff) between walking with and without orthotics during right stance in subjects with flatfeet.

(%diff) With vs. Without Orthotics (flat feet)								
Walking during right stance								
	Sub 1	Sub 3	Sub 4	Sub 5	Sub 6	Sub 7	Sub 9	Sub 10
TA	9.51	-23.13	-28.9	-25.49	3.25	-28.28	-25.28	-----
PL	37.19	9.53	-10.67	-6.04	-16.99	-------	-40.34	33.95
SOL	7.52	-7.99	-24.76	-32.09	-2.14	-17.22	-16.44	27.89
LGAS	30.07	17.21	12.76	-16.95	-23.02	-14.79	-5.12	19.45
MGAS	14.64	-20.89	-27.35	-28.23	15.85	-27.83	-1.616	12.15
VMED	45.19	-7.06	-15.42	2.15	23.19	11.35	-11.08	12.82
VLAT	------	-23.34	-11.02	3.24	-17.75	71.84	36.58	2.30
BFEM	-35.47	10.00	48.85	45.16	-45.25	4.14	-11.60	0.64

Table C.2: EMG percent difference (% diff) between walking with and without orthotics during right stance in subjects without flatfeet.

(%diff) With vs. Without Orthotics (no flat feet)						
Walking during right stance						
	Sub 2	Sub 8	Sub 11	Sub 12	Sub 13	Sub 14
TA	-24.15	-7.34	-4.01	34.32	6.97	16.79
PL	-0.80	-7.76	-11.25	-15.23	-1.07	------
SOL	-11.50	4.22	-6.90	0.44	22.13	76.14
LGAS	-6.57	3.64	7.32	-3.19	3.61	66.59
VMED	-1.35	9.88	55.72	9.91	-7.63	------
MGAS	-39.59	-31.06	-4.05	25.39	-11.76	------
VLAT	-14.65	25.40	44.30	11.94	------	------
BFEM	17.04	-14.33	-8.42	8.85	-11.98	------

Table C.3: EMG percent difference (% diff) between running with and without orthotics during right stance in subjects with flat feet.

(%diff) With vs. Without Orthotics (flat feet) Running during right stance								
	Sub 1	Sub 3	Sub 4	Sub 5	Sub 6	Sub 7	Sub 9	Sub 10
TA	5.03	10.53	22.96	-8.04	-9.83	0.152	-2.24	------
PL	-1.87	-13.62	15.82	-12.39	-8.05	79.21	-30.38	19.92
SOL	-0.48	-4.85	13.82	-13.08	-11.36	-2.58	-20.51	7.49
LGAS	-33.15	33.94	16.26	-12.86	-23.38	-1.14	-0.34	18.21
VMED	-11.44	-55.79	48.42	-3.89	4.51	32.33	10.99	9.35
MGAS	-0.21	-7.22	19.31	-15.00	26.22	8.35	0.612	9.00
VLAT	--------	-8.13	20.75	6.57	-9.65	10.20	27.40	11.02
BFEM	-58.61	0.756	14.01	15.03	-1.84	35.74	-7.40	38.74

Table C.4: EMG percent difference (% diff) between running with and without orthotic conditions during right stance in subjects without flat feet.

(%diff) With vs. Without Orthotics (no flat feet) Running during right stance						
	Sub 2	Sub 8	Sub 11	Sub 12	Sub 13	Sub 14
TA	45.04	-5.15	-21.98	9.18	-10.67	13.57
PL	-13.44	2.44	-4.01	7.97	2.60	62.08
SOL	-22.20	-11.19	-17.10	-1.99	9.15	-39.06
LGAS	-12.20	-15.49	-0.406	-17.91	19.61	71.36
VMED	-26.02	-1.47	0.99	27.75	19.76	-----
MGAS	-20.58	18.93	-17.88	7.82	34.30	------
VLAT	-28.02	-0.95	-15.11	-2.22	-----	------
BFEM	-22.70	-10.79	-16.63	-14.56	7.58	-------

Table C.5: EMG RMS percent difference (% diff) between running with and without orthotics during left stance in subjects with flatfeet.

(%diff) With vs. Without Orthotics (flat feet)								
Running during left stance								
	Sub 1	Sub 3	Sub 4	Sub 5	Sub 6	Sub 7	Sub 9	Sub 10
PL	-8.83	-12.21	10.34	-0.702	0.94	-14.72	-4.36	9.53
LGAS	5.62	-2.07	23.05	-7.20	-19.25	-15.50	-7.11	-3.93
BFEM	-47.98	-8.69	5.60	4.08	7.40	-30.63	48.13	16.79

Table C.6: EMG percent difference (% diff) between the running with and without orthotics during left stance in subjects without flatfeet.

(%diff) With vs. Without Orthotics (no flat feet)						
Running during left stance						
	Sub 2	Sub 8	Sub 11	Sub 12	Sub 13	Sub 14
PL	-23.51	25.48	-31.83	-15.43	52.19	15.77
LGAS	-24.73	8.59	-15.68	3.45	-------	-29.22
BFEM	-1.42	8.66	-16.23	20.37	15.49	7.86

Appendix D
ANOVA Tables

The tables presented below are tables used to display the statistical procedures used for each parameter. ANOVA tables and t-tests (where necessary) are presented as follows.

D1. Walking

D1.1 Kinetics

Table D.1: ANOVA of medio-lateral forces during walking.

FORCE; LS Means (Forces.sta)
Current effect: F(1, 12)=2.0097, p=.18174
Effective hypothesis decomposition

Cell No.	FORCE	Newtons Mean	Newtons Std.Err.	Newtons -95.00%	Newtons +95.00%	N
1	Walking w/o orthotics	94.30963	8.973817	74.75736	113.8619	14
2	Walking w/ orthotics	89.74821	7.075131	74.33282	105.1636	14

Table D.2: ANOVA between groups of medio-lateral forces during walking.

FORCE*Group; LS Means (Forces.sta)
Current effect: F(1, 12)=.54032, p=.47642
Effective hypothesis decomposition

Cell No.	Group	FORCE	Newtons Mean	Newtons Std.Err.	Newtons -95.00%	Newtons +95.00%	N
1	flat feet	Walking w/o orthotics	92.30600	11.74948	66.70607	117.9059	8
2	flat feet	Walking w/ orthotics	90.10975	9.26352	69.92627	110.2932	8
3	no flat feet	Walking w/o orthotics	96.31325	13.56714	66.75300	125.8735	6
4	no flat feet	Walking w/ orthotics	89.38667	10.69659	66.08079	112.6925	6

D1.2 Kinematics

Table D.3: ANOVA of right tibial inclination during walking.

TIBINCL; LS Means (tibial_incl.sta)
Current effect: F(1, 10)=.69693, p=.42331
Effective hypothesis decomposition

Cell No.	TIBINCL	Degrees Mean	Degrees Std.Err.	Degrees -95.00%	Degrees +95.00%	N
1	R tib incl walking w/o orthotics	22.73125	3.017498	16.00784	29.45466	12
2	R tib incl walking w/ orthotics	21.24188	2.938871	14.69366	27.79009	12

Table D.4: ANOVA between groups right tibial inclination during walking.

TIBINCL*Group; LS Means (tibial_incl.sta)
Current effect: F(1, 10)=.38928, p=.54664
Effective hypothesis decomposition

Cell No.	Group	TIBINCL	Degrees Mean	Degrees Std.Err.	Degrees -95.00%	Degrees +95.00%	N
1	flat feet	R tib incl walking w/o orthotics	25.76250	3.484307	17.99898	33.52602	8
2	flat feet	R tib incl walking w/ orthotics	25.38625	3.393516	17.82503	32.94747	8
3	no flat feet	R tib incl walking w/o orthotics	19.70000	4.927554	8.72073	30.67927	4
4	no flat feet	R tib incl walking w/ orthotics	17.09750	4.799156	6.40431	27.79069	4

D1.3 EMG

D1.3.1 Right EMG

Table D.5: ANOVA of the right TA muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=1.4451, p=.25455 Effective hypothesis decomposition					
1	TA w/o orthotics	0.166477	0.016971	0.129125	0.203830	13
2	TA w/ orthotics	0.154269	0.016506	0.117941	0.190598	13

Table D.6: ANOVA between groups of the right TA muscle during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=2.1502, p=.17055 Effective hypothesis decomposition						
1	flat feet	TA w/o orthotics	0.168971	0.023059	0.118219	0.219724	7
2	flat feet	TA w/ orthotics	0.141871	0.022427	0.092511	0.191232	7
3	no flat feet	TA w/o orthotics	0.163983	0.024906	0.109165	0.218802	6
4	no flat feet	TA w/ orthotics	0.166667	0.024224	0.113351	0.219982	6

Table D.7: ANOVA of the right PL muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 10)=.79230, p=.39432 Effective hypothesis decomposition					
1	PL w/o orthotics	0.186047	0.022539	0.135828	0.236266	12
2	PL w/ orthotics	0.172540	0.014892	0.139359	0.205721	12

Table D.8: ANOVA between groups of the right PL muscle during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 10)=.00041, p=.98425 Effective hypothesis decomposition						
1	flat feet	PL w/o orthotics	0.202014	0.029097	0.137182	0.266847	7
2	flat feet	PL w/ orthotics	0.188200	0.019225	0.145364	0.231036	7
3	no flat feet	PL w/o orthotics	0.170080	0.034428	0.093369	0.246791	5
4	no flat feet	PL w/ orthotics	0.156880	0.022748	0.106195	0.207565	5

Table D.9: ANOVA of the right SOL muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 12)=2.8481, p=.11728 Effective hypothesis decomposition					
1	SOL w/o orthotics	0.231787	0.044553	0.134715	0.328860	14
2	SOL w/ orthotics	0.217129	0.043009	0.123421	0.310837	14

Table D.10: ANOVA between groups of the right SOL muscle during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 12)=2.6135, p=.13193 Effective hypothesis decomposition						
1	flat feet	SOL w/o orthotics	0.309725	0.058334	0.182627	0.436823	8
2	flat feet	SOL w/ orthotics	0.281025	0.056312	0.158333	0.403717	8
3	no flat feet	SOL w/o orthotics	0.153850	0.067358	0.007090	0.300610	6
4	no flat feet	SOL w/ orthotics	0.153233	0.065023	0.011560	0.294906	6

Table D.11: ANOVA of the right LGAS muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 12)=.16201, p=.69439 Effective hypothesis decomposition					
1	LGAS w/o orthotics	0.182869	0.032798	0.111408	0.254330	14
2	LGAS w/ orthotics	0.187002	0.032681	0.115796	0.258208	14

Table D.12: ANOVA between groups of the LGAS during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 12)=.00095, p=.97591 Effective hypothesis decomposition						
1	flat feet	LGAS w/o orthotics	0.243288	0.042943	0.149723	0.336852	8
2	flat feet	LGAS w/ orthotics	0.247738	0.042789	0.154507	0.340968	8
3	no flat feet	LGAS w/o orthotics	0.122450	0.049586	0.014411	0.230489	6
4	no flat feet	LGAS w/ orthotics	0.126267	0.049409	0.018614	0.233920	6

Table D.13: ANOVA of the right VMED muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=.45879, p=.51219 Effective hypothesis decomposition					
1	VMED w/o orthotics	0.159095	0.057705	0.032086	0.286104	13
2	VMED w/ orthotics	0.166106	0.053229	0.048950	0.283263	13

Table D.14: ANOVA between groups of the right VMED during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=.03544, p=.85410 Effective hypothesis decomposition						
1	flat feet	VMED w/o orthotics	0.225250	0.071575	0.067715	0.382785	8
2	flat feet	VMED w/ orthotics	0.230312	0.066023	0.084998	0.375627	8
3	no flat feet	VMED w/o orthotics	0.092940	0.090536	-0.106328	0.292208	5
4	no flat feet	VMED w/ orthotics	0.101900	0.083513	-0.081910	0.285710	5

Table D.15: ANOVA of the right MGAS muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=1.7155, p=.21697 Effective hypothesis decomposition					
1	MGAS w/o orthotics	0.197797	0.031861	0.127672	0.267923	13
2	MGAS w/ orthotics	0.178806	0.034049	0.103864	0.253748	13

Table D.16: ANOVA between groups of the right MGAS during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=.03806, p=.84888 Effective hypothesis decomposition						
1	flat feet	MGAS w/o orthotics	0.250975	0.039519	0.163995	0.337955	8
2	flat feet	MGAS w/ orthotics	0.234813	0.042233	0.141859	0.327766	8
3	no flat feet	MGAS w/o orthotics	0.144620	0.049988	0.034598	0.254642	5
4	no flat feet	MGAS w/ orthotics	0.122800	0.053421	0.005222	0.240378	5

Table D.17: ANOVA of the right VLAT muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 9)=.20592, p=.66073 Effective hypothesis decomposition					
1	VLAT w/o orthotics	0.149196	0.036672	0.066238	0.232155	11
2	VLAT w/ orthotics	0.156305	0.029822	0.088844	0.223766	11

Table D.18: ANOVA between groups of the right VLAT muscle during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 9)=.39662, p=.54450 Effective hypothesis decomposition						
1	flat feet	VLAT w/o orthotics	0.156443	0.044228	0.056391	0.256495	7
2	flat feet	VLAT w/ orthotics	0.153686	0.035966	0.072325	0.235047	7
3	no flat feet	VLAT w/o orthotics	0.141950	0.058509	0.009594	0.274306	4
4	no flat feet	VLAT w/ orthotics	0.158925	0.047579	0.051294	0.266556	4

Table D.19: ANOVA of the right BFEM muscle during walking.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=.33137, p=.57644 Effective hypothesis decomposition					
1	BFEM w/o orthotics	0.162440	0.042236	0.069478	0.255402	13
2	BFEM w/ orthotics	0.156056	0.039958	0.068109	0.244004	13

Table D.20: ANOVA between groups of the right BFEM muscle during walking.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right walking2.sta) Current effect: F(1, 11)=.17234, p=.68602 Effective hypothesis decomposition						
1	flat feet	BFEM w/o orthotics	0.171700	0.052388	0.056395	0.287005	8
2	flat feet	BFEM w/ orthotics	0.160713	0.049562	0.051627	0.269798	8
3	no flat feet	BFEM w/o orthotics	0.153180	0.066266	0.007329	0.299031	5
4	no flat feet	BFEM w/ orthotics	0.151400	0.062692	0.013416	0.289384	5

D2. Running

D2.1 Kinetics

Table D.21: ANOVA of medio-lateral forces during running.

Cell No.	FORCE	Newtons Mean	Newtons Std.Err.	Newtons -95.00%	Newtons +95.00%	N
	FORCE; LS Means (Forces.sta) Current effect: F(1, 12)=.09187, p=.76700 Effective hypothesis decomposition					
1	Running w/o orthotics	124.1438	14.83143	91.82891	156.4587	14
2	Running w/ orthotics	122.5910	13.68934	92.76453	152.4175	14

Table D.22: ANOVA between groups of medio-lateral forces during running.

Cell No.	Group	FORCE	Newtons Mean	Newtons Std.Err.	Newtons -95.00%	Newtons +95.00%	N
	FORCE*Group; LS Means (Forces.sta) Current effect: F(1, 12)=.00416, p=.94962 Effective hypothesis decomposition						
1	flat feet	Running w/o orthotics	126.1608	19.41890	83.85059	168.4709	8
2	flat feet	Running w/ orthotics	124.9385	17.92355	85.88644	163.9906	8
3	no flat feet	Running w/o orthotics	122.1269	22.42302	73.27134	170.9825	6
4	no flat feet	Running w/ orthotics	120.2436	20.69633	75.15013	165.3370	6

D2.2 Kinematics

Table D.23: ANOVA of right tibial inclination during running.

	TIBINCL; LS Means (tibial_incl.sta) Current effect: F(1, 11)=2.8005, p=.12240 Effective hypothesis decomposition					
Cell No.	TIBINCL	Degrees Mean	Degrees Std.Err.	Degrees -95.00%	Degrees +95.00%	N
1	R tib incl running w/o orthotics	13.77250	1.905157	9.579278	17.96572	13
2	R tib incl running w/ orthotics	12.73875	1.807783	8.759847	16.71765	13

Table D.24: ANOVA between groups for right tibial inclination during running.

	TIBINCL*Group; LS Means (tibial_incl.sta) Current effect: F(1, 11)=.03542, p=.85415 Effective hypothesis decomposition						
Cell No.	Group	TIBINCL	Degrees Mean	Degrees Std.Err.	Degrees -95.00%	Degrees +95.00%	N
1	flat feet	R tib incl running w/o orthotics	16.55500	2.363056	11.35395	21.75605	8
2	flat feet	R tib incl running w/ orthotics	15.63750	2.242279	10.70228	20.57272	8
3	no flat feet	R tib incl running w/o orthotics	10.99000	2.989056	4.41113	17.56887	5
4	no flat feet	R tib incl running w/ orthotics	9.84000	2.836283	3.59738	16.08262	5

D2.3 EMG

D2.3.1 Left EMG

Table D.25: ANOVA for the left PL muscle during running.

Cell No.	MUSCLE; LS Means (EMG left running2.sta) Current effect: F(1, 12)=.34264, p=.56915 Effective hypothesis decomposition					
	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	PL w/o orthotics	0.271723	0.026376	0.214255	0.329190	14
2	PL w/ orthotics	0.263735	0.024274	0.210846	0.316625	14

Table D.26: ANOVA between groups for the left PL muscle during running.

Cell No.	MUSCLE*Group; LS Means (EMG left running2.sta) Current effect: F(1, 12)=.00091, p=.97638 Effective hypothesis decomposition						
	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	PL w/o orthotics	0.315163	0.034534	0.239920	0.390405	8
2	flat feet	PL w/ orthotics	0.307588	0.031783	0.238339	0.376836	8
3	no flat feet	PL w/o orthotics	0.228283	0.039876	0.141401	0.315166	6
4	no flat feet	PL w/ orthotics	0.219883	0.036699	0.139922	0.299844	6

Table D.27: ANOVA for the left LGAS muscle during running.

Cell No.	MUSCLE; LS Means (EMG left running2.sta) Current effect: F(1, 11)=2.5333, p=.13978 Effective hypothesis decomposition					
	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	LGAS w/o orthotics	0.306525	0.034957	0.229586	0.383464	13
2	LGAS w/ orthotics	0.285285	0.031834	0.215219	0.355351	13

Table D.28: ANOVA between groups for the left LGAS muscle during running.

Cell No.	MUSCLE*Group; LS Means (EMG left running2.sta) Current effect: F(1, 11)=.07854, p=.78448 Effective hypothesis decomposition						
	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	LGAS w/o orthotics	0.405950	0.043359	0.310518	0.501382	8
2	flat feet	LGAS w/ orthotics	0.388450	0.039485	0.301544	0.475356	8
3	no flat feet	LGAS w/o orthotics	0.207100	0.054845	0.086387	0.327813	5
4	no flat feet	LGAS w/ orthotics	0.182120	0.049945	0.072192	0.292048	5

Table D.29: ANOVA of the left BFEM muscle during running.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG left running2.sta) Current effect: F(1, 12)=.00052, p=.98216 Effective hypothesis decomposition					
1	BFEM w/o orthotics	0.313302	0.059295	0.184110	0.442494	14
2	BFEM w/ orthotics	0.312890	0.054837	0.193411	0.432369	14

Table D.30: ANOVA between groups of the left BFEM muscle during running.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG left running2.sta) Current effect: F(1, 12)=.50318, p=.49166 Effective hypothesis decomposition						
1	flat feet	BFEM w/o orthotics	0.426188	0.077635	0.257036	0.595339	8
2	flat feet	BFEM w/ orthotics	0.412963	0.071798	0.256528	0.569397	8
3	no flat feet	BFEM w/o orthotics	0.200417	0.089645	0.005097	0.395736	6
4	no flat feet	BFEM w/ orthotics	0.212817	0.082905	0.032181	0.393452	6

D2.3.2 Right EMG

Table D.31: ANOVA of the right TA muscle during running.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 12)=1.1527, p=.30409 Effective hypothesis decomposition					
1	TA w/o orthotics	0.161642	0.019438	0.119290	0.203993	14
2	TA w/ orthotics	0.171602	0.023962	0.119393	0.223811	14

Table D.32: ANOVA between groups of the right TA muscle during running.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.42852, p=.52507 Effective hypothesis decomposition						
1	flat feet	TA w/o orthotics	0.139800	0.025450	0.084349	0.195251	8
2	flat feet	TA w/ orthotics	0.143687	0.031374	0.075330	0.212045	8
3	no flat feet	TA w/o orthotics	0.183483	0.029387	0.119454	0.247513	6
4	no flat feet	TA w/ orthotics	0.199517	0.036227	0.120584	0.278449	6

Table D.33: ANOVA of the right PL muscle during running.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.69148, p=.42190 Effective hypothesis decomposition					
1	PL w/o orthotics	0.215965	0.022055	0.167911	0.264018	14
2	PL w/ orthotics	0.238758	0.044197	0.142460	0.335056	14

Table D.34: ANOVA between groups of the right PL muscle during running.

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE*Group; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.35334, p=.56327 Effective hypothesis decomposition						
1	flat feet	PL w/o orthotics	0.265863	0.028876	0.202946	0.328779	8
2	flat feet	PL w/ orthotics	0.304950	0.057868	0.178867	0.431033	8
3	no flat feet	PL w/o orthotics	0.166067	0.033344	0.093417	0.238716	6
4	no flat feet	PL w/ orthotics	0.172567	0.066820	0.026978	0.318155	6

Table D.35: ANOVA of the right SOL muscle during running.

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 12)=7.3881, p=.01867 Effective hypothesis decomposition					
1	SOL w/o orthotics	0.323000	0.034709	0.247374	0.398626	14
2	SOL w/ orthotics	0.292702	0.030446	0.226366	0.359038	14

Table D.36: ANOVA between groups of the right SOL muscle during running.

	MUSCLE*Group; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.88487, p=.36542 Effective hypothesis decomposition						
Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	SOL w/o orthotics	0.395200	0.045445	0.296183	0.494217	8
2	flat feet	SOL w/ orthotics	0.375387	0.039863	0.288533	0.462242	8
3	no flat feet	SOL w/o orthotics	0.250800	0.052476	0.136465	0.365135	6
4	no flat feet	SOL w/ orthotics	0.210017	0.046030	0.109726	0.310307	6

Table D.37: ANOVA of the right LGAS muscle during running.

	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.00731, p=.93328 Effective hypothesis decomposition					
Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	LGAS w/o orthotics	0.212779	0.019862	0.169503	0.256055	14
2	LGAS w/ orthotics	0.211658	0.022815	0.161948	0.261369	14

Table D.38: ANOVA between groups of the right LGAS muscle during running.

	MUSCLE*Group; LS Means (EMG right running2.sta) Current effect: F(1, 12)=.04073, p=.84344 Effective hypothesis decomposition						
Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	LGAS w/o orthotics	0.282325	0.026006	0.225663	0.338987	8
2	flat feet	LGAS w/ orthotics	0.283850	0.029872	0.218764	0.348936	8
3	no flat feet	LGAS w/o orthotics	0.143233	0.030029	0.077806	0.208661	6
4	no flat feet	LGAS w/ orthotics	0.139467	0.034494	0.064312	0.214622	6

Table D.39: ANOVA of the right VMED muscle during running.

	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 11)=.06346, p=.80575 Effective hypothesis decomposition					
Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	VMED w/o orthotics	0.240935	0.055805	0.118109	0.363761	13
2	VMED w/ orthotics	0.230541	0.038457	0.145898	0.315185	13

Table D.40: ANOVA between groups of the right VMED muscle during running.

MUSCLE*Group; LS Means (EMG right running2.sta)
Current effect: F(1, 11)=.27138, p=.61273
Effective hypothesis decomposition

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	VMED w/o orthotics	0.296950	0.069218	0.144603	0.449297	8
2	flat feet	VMED w/ orthotics	0.265063	0.047700	0.160075	0.370050	8
3	no flat feet	VMED w/o orthotics	0.184920	0.087554	-0.007786	0.377626	5
4	no flat feet	VMED w/ orthotics	0.196020	0.060336	0.063220	0.328820	5

Table D.41: ANOVA of the right MGAS muscle during running.

MUSCLE; LS Means (EMG right running2.sta)
Current effect: F(1, 11)=.05952, p=.81174
Effective hypothesis decomposition

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	MGAS w/o orthotics	0.250879	0.035670	0.172370	0.329387	13
2	MGAS w/ orthotics	0.253343	0.036700	0.172566	0.334119	13

Table D.42: ANOVA between groups of the right MGAS muscle during running.

MUSCLE*Group; LS Means (EMG right running2.sta)
Current effect: F(1, 11)=.15097, p=.70503
Effective hypothesis decomposition

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	MGAS w/o orthotics	0.308237	0.044243	0.210860	0.405615	8
2	flat feet	MGAS w/ orthotics	0.314625	0.045521	0.214434	0.414816	8
3	no flat feet	MGAS w/o orthotics	0.193520	0.055963	0.070346	0.316694	5
4	no flat feet	MGAS w/ orthotics	0.192060	0.057580	0.065327	0.318793	5

Table D.43: ANOVA of the right VLAT muscle during running.

MUSCLE; LS Means (EMG right running2.sta)
Current effect: F(1, 9)=.00474, p=.94664
Effective hypothesis decomposition

Cell No.	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	VLAT w/o orthotics	0.210943	0.040662	0.118960	0.302926	11
2	VLAT w/ orthotics	0.211521	0.043993	0.112003	0.311039	11

Table D.44: ANOVA between groups of the right VLAT muscle during running.

MUSCLE*Group; LS Means (EMG right running2.sta)
Current effect: F(1, 9)=6.4663, p=.03156
Effective hypothesis decomposition

Cell No.	Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	VLAT w/o orthotics	0.239486	0.049040	0.128550	0.350422	7
2	flat feet	VLAT w/ orthotics	0.261443	0.053057	0.141420	0.381466	7
3	no flat feet	VLAT w/o orthotics	0.182400	0.064874	0.035646	0.329154	4
4	no flat feet	VLAT w/ orthotics	0.161600	0.070188	0.002824	0.320376	4

Table D.45: ANOVA of the right BFEM muscle during running.

Cell No.	MUSCLE; LS Means (EMG right running2.sta) Current effect: F(1, 11)=.00245, p=.96143 Effective hypothesis decomposition MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	BFEM w/o orthotics	0.190856	0.034398	0.115148	0.266565	13
2	BFEM w/ orthotics	0.190280	0.037579	0.107569	0.272991	13

Table D.46: ANOVA between groups of the right BFEM muscle during running.

Cell No.	MUSCLE*Group; LS Means (EMG right running2.sta) Current effect: F(1, 11)=3.2395, p=.09934 Effective hypothesis decomposition Group	MUSCLE	Volts Mean	Volts Std.Err.	Volts -95.00%	Volts +95.00%	N
1	flat feet	BFEM w/o orthotics	0.192012	0.042665	0.098108	0.285917	8
2	flat feet	BFEM w/ orthotics	0.212400	0.046611	0.109810	0.314990	8
3	no flat feet	BFEM w/o orthotics	0.189700	0.053967	0.070919	0.308481	5
4	no flat feet	BFEM w/ orthotics	0.168160	0.058959	0.038393	0.297927	5

Appendix E
Paired T-test tables

The tables presented below are tables used to display the paired t-tests used for each parameter.

E1. Walking

E1.1 Kinematics

Table E.1: Paired t-test for tibial inclination without orthotics during walking.

Variable	T-test for Dependent Samples (tibial_incl.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
R tib incl walking w/o orthotics	25.25455	8.75854						
L tib incl walking w/o orthotics	21.17182	12.49138	11	4.082727	10.47559	1.292612	10	0.225213

E1.2 EMG

Table E.2: Paired t-test of the TA muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right TA w/o orthotics	0.166669	0.058468						
Left TA w/o orthotics	0.141662	0.072834	13	0.025008	0.063951	1.409922	12	0.183952

Table E.3: Paired t-test of the PL muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right PL w/o orthotics	0.188708	0.075221						
Left PL w/o orthotics	0.194283	0.065676	12	-0.005575	0.043668	-0.442255	11	0.666874

Table E.4: Paired t-test of the SOL muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right SOL w/o orthotics	0.241875	0.192893						
Left SOL w/o orthotics	0.200658	0.126765	12	0.041217	0.103776	1.375830	11	0.196240

Table E.5: Paired t-test of the LGAS muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right LGAS w/o orthotics	0.195154	0.136829						
Left LGAS w/o orthotics	0.215046	0.163628	13	-0.019892	0.102689	-0.698448	12	0.4982

Table E.6: Paired t-test of the VMED muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right VMED w/o orthotics	0.177210	0.233840						
Left VMED w/o orthotics	0.202150	0.312343	10	-0.024940	0.126539	-0.623264	9	0.5485

Table E.7: Paired t-test of the MGAS muscle without orthotics during walking.

Variable	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right MGAS w/o orthotics	0.209458	0.125110						
Left MGAS w/o orthotics	0.215692	0.215594	12	-0.006233	0.146718	-0.147173	11	0.885

Table E.8: Paired t-test of the VLAT muscle without orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right VLAT w/o orthotics	0.153111	0.123140						
Left VLAT w/o orthotics	0.251578	0.365077	9	-0.098467	0.293804	-1.00543	8	0.344127

Table E.9: Paired t-test of the BFEM muscle without orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right BFEM w/o orthotics	0.164577	0.142177						
Left BFEM w/o orthotics	0.250577	0.222057	13	-0.08600C	0.22391E	-1.3848C	12	0.19132

Table E.10: Paired t-test of the TA muscle with orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right TA w/ orthotics	0.153315	0.058248						
Left TA w/ orthotics	0.144331	0.052199	13	0.008985	0.045743	0.708178	12	0.492362

Table E.11: Paired t-test of the PL muscle with orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right PL w/ orthotics	0.168577	0.054370						
Left PL w/ orthotics	0.188254	0.064325	13	-0.019677	0.051228	-1.38492	12	0.191284

Table E.12: Paired t-test of the SOL muscle with orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right SOL w/ orthotics	0.222808	0.180756						
Left SOL w/ orthotics	0.194692	0.118868	12	0.028117	0.100404	0.970072	11	0.352858

Table E.13: Paired t-test of the LGAS muscle with orthotics during walking.

	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right LGAS w/ orthotics	0.201292	0.135592						
Left LGAS w/ orthotics	0.203485	0.150499	13	-0.002192	0.057046	-0.138564	12	0.89209

Table E.14: Paired t-test of the VMED muscle with orthotics during walking.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Right VMED w/ orthotics	0.171990	0.213506						
Left VMED w/ orthotics	0.221300	0.343187	10	-0.049310	0.149831	-1.04072	9	0.325157

Table E.15: Paired t-test of the MGAS muscle with orthotics during walking.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Right MGAS w/ orthotics	0.194633	0.132887						
Left MGAS w/ orthotics	0.222567	0.220871	12	-0.027933	0.138829	-0.697001	11	0.50026

Table E.16: Paired t-test of the VLAT muscle with orthotics during walking.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Right VLAT w/ orthotics	0.152711	0.100309						
Left VLAT w/ orthotics	0.265767	0.351345	9	-0.113056	0.301533	-1.12481	8	0.293284

Table E.17: Paired t-test of the BFEM muscle with orthotics during walking.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG walking2.sta) Marked differences are significant at p < .05000							
Right BFEM w/ orthotics	0.157131	0.134298						
Left BFEM w/ orthotics	0.235969	0.237969	13	-0.078838	0.208767	-1.36159	12	0.198343

E2. Running

E2.1 Kinematics

Table E.18: Paired t-test for tibial inclination without orthotics during running.

Variable	T-test for Dependent Samples (tibial_incl.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
R tib incl running w/o orthotics	14.37308	6.929744						
L tib incl running w/o orthotics	12.47077	7.030335	13	1.902308	4.997596	1.372433	12	0.195036

Table E.19: Paired t-test for tibial inclination with orthotics during running.

Variable	T-test for Dependent Samples (tibial_incl.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
R tib incl running w/ orthotics	13.40769	6.744541						
L tib incl running w/ orthotics	11.74154	6.149772	13	1.666154	3.886480	1.545718	12	0.148126

Table E.20: Paired t-test for tibial inclination with orthotics during running.

Variable	T-test for Dependent Samples (tibial_incl.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
R tib incl walking w/ orthotics	25.21250	7.00312						
L tib incl walking w/ orthotics	24.32333	11.69570	12	0.889167	8.205075	0.375397	11	0.714503

E2.2 EMG

Table E.21: Paired t-test of the TA muscle without orthotics during running.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Right TA w/o orthotics	0.158521	0.072707						
Left TA w/o orthotics	0.169321	0.079687	14	-0.010800	0.049028	-0.824223	13	0.424677

Table E.22: Paired t-test of the PL muscle without orthotics during running.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Right PL w/o orthotics	0.223093	0.093724						
Left PL w/o orthotics	0.277929	0.103911	14	-0.054836	0.052852	-3.88209	13	0.001889

Table E.23: Paired t-test of the SOL muscle without orthotics during running.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Right SOL w/o orthotics	0.346808	0.151885						
Left SOL w/o orthotics	0.306442	0.127030	12	0.040367	0.082080	1.703641	11	0.116498

Table E.24: Paired t-test of the LGAS muscle without orthotics during running.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Right LGAS w/o orthotics	0.233385	0.095975						
Left LGAS w/o orthotics	0.329469	0.154678	13	-0.096085	0.086439	-4.00787	12	0.00173

Table E.25: Paired t-test of the VMED muscle without orthotics during running.

Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Right VMED w/o orthotics	0.273973	0.207350						
Left VMED w/o orthotics	0.248064	0.317682	11	0.025909	0.169334	0.507462	10	0.622834

Table E.26: Paired t-test of the MGAS muscle without orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right MGAS w/o orthotics	0.264115	0.133150						
Left MGAS w/o orthotics	0.275400	0.214339	13	-0.011285	0.189810	-0.214358	12	0.8338

Table E.27: Paired t-test of the VLAT muscle without orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right VLAT w/o orthotics	0.218780	0.133252						
Left VLAT w/o orthotics	0.353420	0.368602	10	-0.134640	0.317561	-1.34075	9	0.21285

Table E.28: Paired t-test of the BFEM muscle without orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right BFEM w/o orthotics	0.191123	0.115543						
Left BFEM w/o orthotics	0.340246	0.246994	13	-0.149123	0.234646	-2.29141	12	0.04082

Table E.29: Paired t-test of the TA muscle with orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right TA w/ orthotics	0.167614	0.089949						
Left TA w/ orthotics	0.162071	0.066145	14	0.005543	0.071076	0.291791	13	0.775051

Table E.30: Paired t-test of the PL muscle with orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right PL w/ orthotics	0.248214	0.171321						
Left PL w/ orthotics	0.270000	0.097407	14	-0.021786	0.115223	-0.707450	13	0.491776

Table E.31: Paired t-test of the SOL muscle with orthotics during running.

Variable	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right SOL w/ orthotics	0.314258	0.146849						
Left SOL w/ orthotics	0.285842	0.109645	12	0.028417	0.076265	1.290742	11	0.223262

Table E.32: Paired t-test of the LGAS muscle with orthotics during running.

	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right LGAS w/ orthotics	0.231323	0.108483						
Left LGAS w/ orthotics	0.309092	0.149496	13	-0.077769	0.080377	-3.48857	12	0.004475

Table E.33: Paired t-test of the VMED muscle with orthotics during running.

	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right VMED w/ orthotics	0.252182	0.141823						
Left VMED w/ orthotics	0.202564	0.240062	11	0.049618	0.228149	0.721306	10	0.487243

Table E.34: Paired t-test of the MGAS muscle with orthotics during running.

	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right MGAS w/ orthotics	0.267485	0.138014						
Left MGAS w/ orthotics	0.289215	0.218618	13	-0.021731	0.189036	-0.414478	12	0.68583

Table E.35: Paired t-test of the VLAT muscle with orthotics during running.

	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right VLAT w/ orthotics	0.231940	0.148185						
Left VLAT w/ orthotics	0.298080	0.242175	10	-0.066140	0.183317	-1.14093	9	0.283347

Table E.36: Paired t-test of the BFEM muscle with orthotics during running.

	T-test for Dependent Samples (EMG running2.sta) Marked differences are significant at p < .05000							
Variable	Mean	Std.Dv.	N	Diff.	Std.Dv. Diff.	t	df	p
Right BFEM w/ orthotics	0.195385	0.128196						
Left BFEM w/ orthotics	0.336685	0.226532	13	-0.141300	0.215699	-2.36192	12	0.035925